Universidade de Aveiro Departamento de Engenharia Mecânica
2018

Luís Miguel Matos Varela

Desenvolvimento de um dispositivo de proteção para impactos na cabeça

Development of a head impact protection device

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Dissertação apresentada à Universidade de Aveiro para cumprimento dos requisitos necessários à obtenção do grau de Mestre em Engenharia Mecânica, realizada sob orientação científica do Doutor Ricardo José Alves de Sousa, Professor Auxiliar do Departamento de Engenharia Mecânica da Universidade de Aveiro e de Fábio António Oliveira Fernandes, Professor Auxiliar Convidado do Departamento de Engenharia Mecânica da Universidade de Aveiro.

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Palavras-chave absorção de energia, caracterização de material, cérebro, cortiça, impacto, desporto, lesões cerebrais, materiais celulares, materiais naturais, equipamento de proteção, simulação numérica

Resumo Hoje em dia o número de praticantes de desportos coletivos envolvendo contacto físico tem aumentado. Para desportos onde o uso de equipamento de proteção ao nivel da cabeça não é obrigatório, alguns desportistas usam uma fita para que se possam sentir mais seguros. No mercado existem várias soluções para esse efeito, feitas com espumas sintéticas que garantem um nivel de proteção proporcional às carateristicas mecânicas das mesmas. A cortiça é um material celular natural capaz de suster quantidades consideráveis de energia. Estas características tornam este material ideal para determinadas aplicações como a proteção de impactos. Neste trabalho, é estudado qual o aglomerado de cortiça, com a melhor densidade e por consequência com a melhores propriedades para incorporar numa fita de proteção da cabeça. Para o efeito, foram utilizadas 3 fitas existentes no mercado de modo a concluir em que nivel se encontrava a cortiça em relação às espumas destes equipamentos. Posteriormente, foi criado um modelo de fita que após validado foi variado o seu material e analisado o risco de lesão. Os resultados desta análise mostraram que o aglomerado de cortiça selecionado consegue ter a mesma e em algumas situações, melhor performance do que as espumas sintéticas encontradas nestas fitas tidas como referência.

Keywords headband, energy absorption, material characterization, brain, cork, impact, sport, brain injury, cellular materials, natural materials, protective equipment, numerical simulation

Abstract Nowadays, the number of people that practise team sports involving physical contact has increased. For sports where the use of head protective equipment is not mandatory, some sportsmen use headbands to feel safer. In the market, there are several solutions for this purpose made by synthetic foams that guarantee a level of protection proportional to its mechanical properties. Cork is a natural cellular material capable of absorbing amounts of energy. These characteristics make this material ideal for certain applications such as impact protection. In this work, the cork agglomerate with the best density and properties to incorporate into a protective headband is studied. With this goal, 3 headbands at the market were used in order to conclude on which level the cork was comparing to their foams. Finally, a headband model was created. After being validated, its material was varied and, for each one, the risk of injury was analysed. The results of these analysis showed that the selected agglomerated cork can have the same and, in some situations, better performance than the foams found in these reference devices.

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List of Acronyms

AE	Athletic Exposures
AIS	Abbreviated Injury Scale
BV	Bridging Veins
CT	Computed Tomography
CTE	Chronic Traumatic Encephalopathy
EPP	Expanded Polypropylene
EVA	Ethylene-Vinyl Acetate
FEHM	Finite Element Head Model
GCS	Glasgow Coma Scale
HIC	Head Injury Criteria
HIP	Head Impact Power
MRI	Magnetic Resonance Imaging
MTBI	Mild Traumatic Brain Injury
NFL	National Football League
NOCSAE	National Operating Committee on Standards for Athletic Equipment
PE	Polyethylene
PLA	Peak of Linear Acceleration
PTA	Posttraumatic Amnesia
PU	Polyurethane
SDH	Subdural Haematoma
SI	Severity Index
SSS	Superior Sagittal Sinus
TBI	Traumatic Brain Injury
VN	Vinyl Nitrile
WSTC	Wayne State Tolerance Curve
YEAHM	Yet Another Head Model

List of Symbols

K_0	Bulk modulus
A_c	Compressed area
ρ	Density
$\overline{I}_1$	Deviatoric strain invariant
$\overline{\lambda_i}$	Deviatoric stretches
J^{el}	Elastic volume ratio
F	Force
L_0	Initial thickness
C_{10}, C_{01}, α_1, D_1	Material parameters
ϵ	Nominal strain
ν	Poisson's ratio
N	polynomial order
λ_i	Principal stretches
g	Relaxation coefficient
τ	Relaxation time
μ	Shear modulus
U	Strain energy
σ	Stress
C_{i0}, D_i	Temperature-dependent parameter
J^{th}	Thermal volume ratio
L	Thickness in the moment
E	Young's modulus

Chapter 1

Introduction

In this first chapter, a brief introduction is made. The scope, motivation and main objectives of this research are presented. In addition, a reading guide with information about each chapter is also provided.

In each country, there is a popular sport, which leads to a huge number of participants. Figure 1.1 presents which sport is the most popular in each country.

Figure 1.1: The most popular sport in each country [1].

One example of this trend is the number of players registered in the Association of soccer from Aveiro, the third biggest association in Portugal. In the Table 1.1 it is presented the huge number of soccer players in teams from Aveiro.

Table 1.1: Number of registered players in the Association of soccer from Aveiro [2].

	Gender	**2010/2011**	**2011/2012**	**2012/2013**	**2013/2014**
Senior	Male	2081	1902	1669	1797
	Female	146	133	99	98
Under-19	Male	1248	1149	1146	1215
	Female	74	79	57	69
Under-17	Male	1471	1478	1592	1553
	Female	0	1	0	61
Under-15	Both	1749	1721	1776	1862
Under-13	Both	1763	1876	1866	1832
Under-11	Both	1763	1876	1844	1744
Under-9	Both	1227	1306	1169	1204
Under-7	Both	99	130	174	159

A second example is the number of ice hockey players in some countries, that like soccer the value is high. In the next table it is presented the top 5 countries with the biggest number of registered ice hockey players, including male, female and junior, provided by the respective countries' federations [3], [4].

Table 1.2: Top 5 countries with the biggest number of registered hockey players

Country	**Players**	**% of population**
Canada	631,295	1.724%
United States	555,935	0.170%
Czech Republic	113,425	1.075%
Russia	105,059	0.073%
Finland	76,387	1.379%

The third example reinforces the information of the Figure 1.1 because the Figure 1.2 presents the number of athletes on each sport in the high school of the US in the season 2015/2016. As was predicted the number of football players was the highest with 1085272 players.

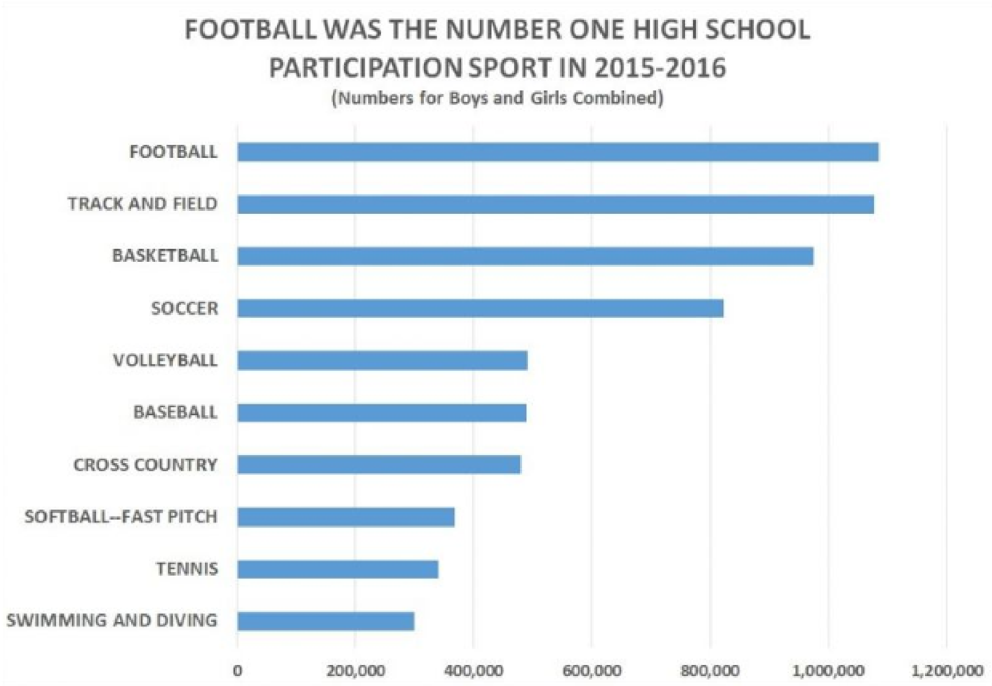


Figure 1.2: Number of high School participants in each sport in 2015/2016 [5].

Soccer, ice hockey, and American football are the most popular sports in the world but also where the contact between players is very frequent. Due to that fact, head injuries started to be a big issue in its routine.

Head injuries had been a worrisome issue in the sporting world, not just because of the injury itself, but also due to the number of sports players that have grown as was saw before.

Concussion is one of the most common head injuries and this worries parents, coaches, players, physicians, etc. Tables 1.3 and 1.4 present some studies results about concussion having in consideration different sport levels, gender and institutions. The results are measured in concussions per 1000 Athletic Exposures (AEs - is defined as one athlete participating in one practice or game where he or she is exposed to the possibility of athletic injury).

Table 1.3: Concussion rates in different sports [6].

Sport	Statistic	Reference
Football	High school incident estimated 0.48/1000 AEs and reported 1.03/1000 AEs.	[7], [8]
	College incident estimated 0.52/1000 AEs and reported 0.81/1000 AEs.	[9], [10]
	NFL incident estimated 4.51/1000 AEs.	[11]
	Multisport studies ranged from 0.33 to 0.64/1000 AEs in high school.	[12], [13]
Rugby	Incidence in male nonprofessional rugby players was 7.97/1000 player hours.	[14]
	Reported incident of 14% per 20 hours season in experienced rugby players.	[15]
	Community rugby club player incident 1.8/1000 hours.	[16]
	Annual Rugby concussions between 4% and 14% at school level and between 3% and 23% adult level.	[17]
Ice Hockey	Concussion ranged from 21.52/1000 AEs in junior hockey to 1.55/1000 AEs in collegiate players.	[18], [19]
	Reported lower incidence in collegiate men at 0.72/1000 AEs than women at 0.82/ 1000 AEs.	[20]
	Reported of 21.52/1000 AEs in junior hockey is 7 times higher than previously reported in the NHL.	[18]
	Multisport studies incidence ranged from 0.41/1000 in collegiate male hockey to 0.54/1000 AEs in high school males.	[13]
	Reported collegiate female hockey incidence at 0.91/1000 AEs.	[21]
Lacrosse	Reported an overall incidence in collegiate men of 1.08/1000 AEs and 0.52/1000 AEs in women.	[22], [23]
	Concussion was 8.6% of all game injury in men and 9.4% in women.	
	Reported an incidence in high school males of 0.28/1000 AEs and 0.21/1000 AEs in females.	[24]
	Reported an incidence in college of 0.87/1000 AEs and females at 0.32/1000 AEs.	[24]
	Multisport study incidences ranged from 0.26/1000 AEs in collegiate males to 0.40/1000 AEs in high school males.	[13], [21]
	Concussion spanned from 0.25/1000 AEs in collegiate females to 0.35/1000 AEs in high school females.	[13], [21]
Soccer	Concussion in the multisport studies ranged from 0.13/1000 AEs in high school females to 0.41/1000 AEs in college females.	[12], [21]
	Male incidence spanned from high school at 0.17 to college at 0.49/1000 AEs.	[25], [26]

Table 1.4: Frequency, distribution, and rates of select injuries for games and practices combined for 15 sports, 1988 - 1989 to 2003 - 2004 [21].

	Percentage all injuries (%)	**Injury Rate per 1000 Athletic Exposures**	**95% Confidence Interval**
Men's baseball	2.5	0.07	0.06, 0.08
Men's basketball	3.2	0.16	0.14, 0.17
Women's basketball	4.7	0.22	0.20, 0.17
Women's field hockey	3.9	0.18	0.15, 0.21
Men's football	6.0	0.37	0.36, 0.38
Women's gymnastics	2.3	0.16	0.12, 0.20
Men's ice hockey	7.9	0.41	0.37, 0.44
Women's ice hockey	18.3	0.91	0.71, 1.11
Men's lacrosse	5.6	0.25	0.23, 0.29
Women's lacrosse	6.3	0.25	0.22, 0.29
Men's soccer	3.9	0.28	0.25, 0.3
Women's soccer	5.3	0.41	0.38, 0.44
Women's softball	4.3	0.14	0.12, 0.16
Men's wrestling	3.3	0.25	0.22, 0.27
Men's spring football	5.6	0.54	0.5, 0.58
Total concussions	5.0	0.28	0.27, 0.28

In these two tables, there are some values that change despite referring to the same sport. This could happen due to the different target group of each study. However, the combination of both shows the impact of concussion in the sport's world.

If we look at young athletes, they report higher incidence despite the small risk factors (less exposure and fewer cumulative concussions) [8], [25].

High school players may be at greater risk because teams run more and pass less than in college and players are less skilled in tackling and blocking techniques. "Immature neurological, vascular, and musculoskeletal physiology combined with less experience, training, and tackling skill may result in more absolute force to the brain per hit and result in higher incidence" [7]. However, older athletes may resist more to report injury due to the scholarship, scouting or progression to a professional status that can end or regress [27].

In the sex-comparable sports, female incidence is higher than male [12], [13], [21], [25], [26], [28], [29], excepting, in some studies about lacrosse, where concussion were more frequent in males, [13], [21], [23], [24], [26]. Player contact was the cause of concussion for 72% of men but 41% of women [20]. Studies have speculated that females have more concussions because of intrinsic differences between the sexes, such as height, weight, head and neck size, or strength [27], [30].

Another important factor in the concussion mechanism is the location of the head impact. Some studies concluded that there are some locals on the head that are more common to be injured than others like is presented in Table 1.5.

Table 1.5: Annual Concussion Counts and Rates^a (and 95% CIs) Among US High School Athletes, by Impact Direction, High School Sports-Related Injury Surveillance Study, United States, 2008/2009 to 2012/2013 [31].

School Year	**Impact Location**					**Total**
	Back	**Front**	**Side**	**Top**	**Unknown^b**	
2008/2009						
n(%)	21 (5.9)	146 (41.0)	68 (19.1)	21 (5.9)	100 (28.1)	365(100)
Rate	0.4	2.5	1.2	0.4	1.7	6.1
(95% CI)	(0.2-0.5)	(2.1-2.9)	(0.9-1.4)	(0.2-0.5)	(1.4-2.0)	(5.4-6.7)
2009/2010						
n (%)	22 (4.9)	189 (42.5)	106 (23.8)	29 (6.5)	99 (22.3)	445 (100)
Rate	0.5	4.0	2.2	0.6	2.1	7.5
(95% CI)	(0.3-0.7)	(3.4-4.5)	(1.8-2.7)	(0.4-0.8)	(1.7-2.5)	(6.7-8.3)
2010/2011						
n (%)	35 (7.5)	219 (47.0)	97 (20.8)	19 (4.1)	96 (20.6)	466 (100)
Rate	0.7	4.6	2.0	0.4	2.0	7.5
(95% CI)	(0.5-1.0)	(4.0-5.2)	(1.6-2.5)	(0.2-0.6)	(1.6-2.4)	(6.7-8.3)
2011/2012						
n (%)	26 (5.1)	235 (46.3)	114 (22.4)	33 (6.5)	100 (19.7)	508 (100)
Rate	0.6	5.0	2.4	0.7	2.1	8.4
(95% CI)	(0.3-0.8)	(4.4-5.6)	(2.0-2.9)	(0.5-0.9)	(1.7-2.5)	(7.6-9.2)
2012/2013						
n (%)	39 (5.2)	341 (45.4)	179 (23.8)	36 (4.8)	156 (20.8)	751 (100)
Rate	0.8	6.7	3.5	0.7	3.1	9.2
(95% CI)	(0.5-1.0)	(6.0-7.4)	(3.0-4.0)	(0.5-0.9)	(2.6-3.5)	(8.3-10.1)
Total						
n (%)	143 (5.7)	1130 (44.7)	564 (22.3)	138 (5.5)	551 (21.8)	2526 (100)
Rate	0.6	4.5	2.2	0.5	2.2	10.0
(95% CI)	(0.5-0.7)	(4.2-4.8)	(2.1-2.4)	(0.5-0.6)	(2.0-2.4)	(9.6-10.4)

a: Rates per 10 000 AEs.

b: AT stated that he or she did not know impact location, or AT did not report impact location (ie, missing).

Most concussions, in Table 1.5, caused by football player-to-player collisions were from front-of-the-head impacts (44.7%), followed by the side-of-the-head (22.3%), back-of-the-head (5.7%), and top-of-the-head (5.5%) impacts. An additional 551 concussions (21.8%) had an unknown impact location. The highest rate of concussions resulting from player-to-player collisions was in competition [31].

Others go a little further and try to achieve to the same objective but on each position in a football game [32]. In the Figure 1.3, there are the final results of this study.

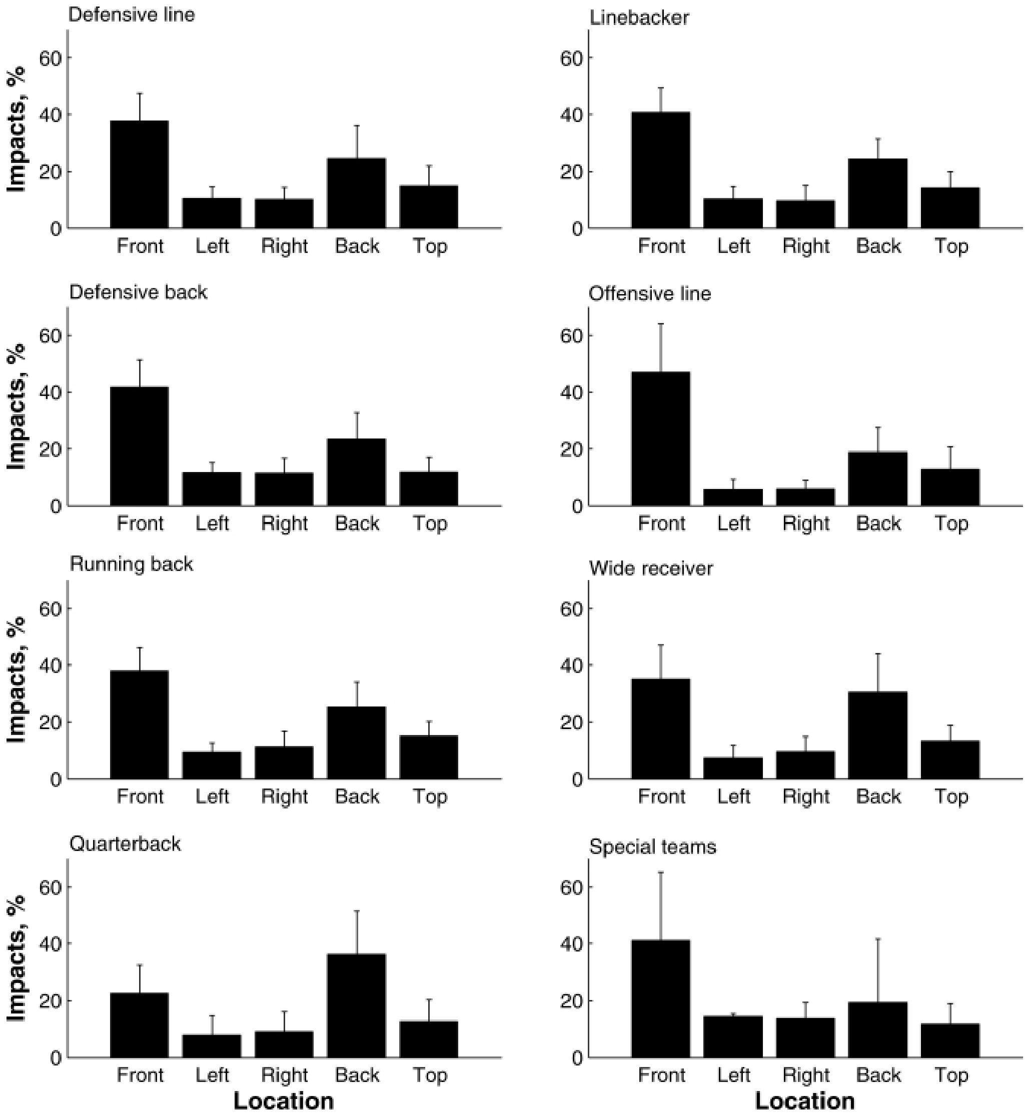


Figure 1.3: Percentage of season head impacts at each helmet location (front, left, right, back, top) [32].

The highest percentage of impacts occurred to the front of the helmet. The back of the helmet received the second highest percentage of impacts and there is no difference between impacts to the left and to the right side [32].

As in the previous study, the results showed that the front of the head is the location that suffers most of the impacts.

1.1 Motivation

Nowadays, more and more are the worries about the seriousness of head injuries that players of soccer, rugby, American football, hockey and others can suffer due to the big number of impacts during the games. Besides that, kids start to play these sports early, that became a more dangerous situation.

As will be presented in section 2.2, there are some alternatives to the classic helmets that could be used in sports where head protection is not required and a helmet would decrease the players performance. One of them is the headband, that covers around 50-55% of the head, made by synthetic cellular materials with the objective to reduce impact force. The majority of these liners used in these devices are able to absorb reasonable amounts of energy by deforming permanently. Under compressive loading, cellular materials can undergo large strain deformation while maintaining its low-stress level almost constant before the material's densification, which allows them to absorb large amounts of energy [33], [34].

Following the United Nation's 2030 agenda of the Sustainable Development goals signed in September of 2015, there is some pressure in the producers to provide eco-friendly alternatives to the current market solutions. Natural materials have been showing a tremendous potential to replace the synthetic ones in a great variety of engineering and design applications. In a society continuously searching for new environmentally friendly and sustainable resources, a material as cork would be a great substitute for these synthetic materials.

Cork is a natural cellular material capable of absorbing great amounts of energy and recovers almost entirely after deformation, which is a desired characteristic in multi-impact applications [33], [35]. In addition, regarding cork agglomerates, there is the possibility to change its density changing its mechanical properties and thus, its behaviour under impact [36]. These features make cork an ideal substitute for synthetic cellular materials [35], [37].

As there are no specific standards for headbands, the producers start to test it in accordance with helmet and headgear standards for sport, trying to show the impact protection of their product. ASTM F2439 (standard specification for headgear used in soccer) and ASTM F1045 (standard performance specification for ice hockey helmets) are two examples follwed for the producers of the ForceField headband in their tests. The only rule that this device must follow is the Law 4 of FIFA which allow the players to use headbands if it was made by soft and lightweight padded material: "Non-dangerous protective equipment, for example headgear, facemasks and knee and arm protectors made of soft, lightweight padded material is permitted." [38].

Over the years in order to obtain the properties and behaviour of the head components, some experimental tests have been made in human corpses and animals. Besides that, with the increasing computing power and advances in computational modelling, led the researchers to try to model some parts of the human body including the head [39]. Several finite element head models (FEHM) have been develop during the last decade [40], [41], [42], [43], [44], [45].

Head model is a great tool to develop some important aspects about headbands and others head protective devices like geometry and dimensions. Once the FEHM is validated, it can be used to optimize the device from a biomechanical point of view [46], [47]. These models are a better approach to the real human head than the rigid headforms and save greats amounts of resources, such as material. It is a more flexible and cheaper procedure. This biomechanical criterion is based on the proposed head level injury predictors [35]. This allows a further accurate, computational-based prediction of the brain injury, relating it to the medical investigations observed in autopsy or injured individuals.

The research presented in this thesis intends to assess agglomerated cork as an energy absorbing material in headbands. Three types of agglomerates and three headbands were tested experimentally. After its characterization, these experiments were simulated in order to assess the validity of the constitutive models used. The best agglomerated cork was selected and saw its potential. Finite element headband model was also developed helping to see if the cork headband model was able to pass in the same standards used to test the headbands already in the market. Then, it was performed a head injury study with a developed FEHM, based on biomechanical criteria. In the end, it was evaluated the applicability of agglomerated cork as an energy absorption liner in headbands.

1.2 Objectives

The main goal of this thesis is to analyse the applicability of agglomerated cork as energy absorption liner in personal safety gear, more specifically in headbands. In the scope of this work, several stages had to be covered:

- Study the advantages of the new headband in terms of peak acceleration reduction and Head Injury Criterion (HIC).
- Discuss the best material solution for the headband in terms of agglomerated cork and, compare its potential with the ones already in the market.
- Study and validate the headband in real situations to evaluate the brain injuries with computational simulation.

The work scenario will be the sport world where the range of head impact events are very diversified. In addition, cork application is not limited to headbands and has the potential to be applied in other types of personal safety gear or even in other applications as it was seen in previous studies [48].

1.3 Reading Guide

This thesis is divided in 5 chapters. In order to provide to the reader a practical reading guide, a small description of all chapters and their contents is performed.

Chapter 1 - Introduction
This chapter presents a brief introduction along with the motivation for this thesis. A summary of the main objectives of this work is also presented. In addition, a reading guide is provided, giving a brief summary of each chapter.

Chapter 2 - State-of-the-art
This chapter presents the state-of-the-art relative to the topics covered in this thesis. Explains some of the most common head injuries in sports but also what is done when it is diagnosed or suspicious. Some head injury mechanisms and head injury criteria, as well as the associated thresholds proposed by several studies in the literature are presented.

It was also made a review to all the head protection devices in sports focusing on the different types existing today and the materials typically used.

Chapter 3 - Material Selection
This chapter describes the procedures and methods used to characterize the mechanical behaviour of synthetic foams and the agglomerated cork involved in this work. The experimental characterisation of quasi-static and dynamic as well as numerical simulations are performed in order to validate the constitutive laws and mechanical properties of all the materials.

Chapter 4 - New Cork-based Headband
Creation and evaluation of a headband model made by all the materials analysed in the previous chapter. In this chapter, was performed a impact test simulation with a headform and the headband model in order to access to the linear acceleration in the center of mass of the headform. These results help to made a direct comparison between liners made of synthetic foams and cork agglomerates.

Biomechanical evaluation with the YEAHM was made with the previous results in order to verify if agglomerated cork liners are an alternative to the ones made of synthetic foams. Additionally, with the thresholds of the first chapter was possible to see which head injury was associated to each impact and material.

Chapter 5 - Conclusions and future works
This chapter presents the general conclusions and discuss the results obtained in this work. The competences acquired along the development of this thesis are also discuss. In addition, some suggestions and ideas to implement in future works are listed.

Chapter 2

State-of-the-art

This chapter presents the state-of-the-art relative to the topics covered in this thesis. It presents some head injuries that can be observed in sports as well as all its assessment and recovery process. This previous literature review covered injury mechanisms, criteria and its thresholds that will be useful in next chapters. In the end, the range of head protection solutions are reviewed and known the materials used on them.

2.1 Sports related head injuries

2.1.1 Traumatic Brain Injury (TBI)

Traumatic brain injury (TBI) is an important cause of morbidity and mortality, accounting for more than 1.4 million annual cases in the United States and an estimated 10 million cases globally [49]. Over the years this problem has been studied in order to establish some guidelines as well as an efficient trauma system with the help of advances in critical care. It has contributed to improved outcomes after TBI [50], [51]. The understanding of the pathophysiology of TBI has also improved helping to refine the acute clinical management of patients.

TBI is an acquired brain injury that is defined by an alteration in brain function, or other evidence in the brain pathology, caused by an external force (bump, blow, jolt to the head, or a penetrating head injury) [52].

There is a range of different possibles signs and symptoms related to TBI [53]:

- Period of loss or a decreased level of consciousness;
- Loss of memory for events immediately before or after the injury;
- Neurologic deficits (weakness, loss of balance, change in vision, sensory loss, etc);

- Alteration in mental state at the time of the injury (confusion, disorientation, slowed thinking, etc)

The type and the duration of them can help to identify which or how severe is the TBI. In the Table 2.1, it is presented the classification of a TBI based on clinical severity and duration of symptoms [54], and characteristics and location of the injury [55].

Table 2.1: Clinical Severity and Duration of Symptoms and characteristics and location of injury. [54], [55].

Clinical Severity and Duration of Symptoms
Concussion or cerebral contusion
- No loss of consciousness or loss of consciousness for less than 6 hours.
- No or mild memory deficit.
- Minutes to hours of posttraumatic amnesia.
- No or mild motor deficits.
Mild
- Loss of consciousness lasting for 6 to 24 hours.
- Mild to moderate memory deficit.
- Hours of posttraumatic amnesia.
- Mild motor deficit.
Moderate
- Loss of consciousness for more than 24 hours.
- Moderate memory deficit.
- Days of posttraumatic amnesia.
- Moderate motor deficit.
Severe
- Loss of consciousness lasting for days to weeks.
- Severe memory deficit.
- Weeks of posttraumatic amnesia.
- Severe motor deficit.
Characteristics and Location of Injury
- Primary or secondary TBI.
- Local or diffuse TBI.

The assessment of the TBI is made using an injury severity score, the most common in this type of situations, the Glasgow Coma Scale (GCS). It consists in 3 categories (eye opening response, verbal response, and motor response), assessing for impaired consciousness and coma when administered in the first 24 hours following the injury. The score range between 3 and 15, where higher scores indicate better function [56]. In the Table 2.2 the scale system is presented as well as the score for some injuries.

However, the result of GCS can be wrong if the manifestation of the symptoms were due to drugs, alcohol, or medications; caused by other injuries or treatment for other injuries (systemic injuries, facial injuries, or intubation); caused by other problems (psychological trauma, language barrier, or coexisting medical conditions); or caused by penetrating craniocerebral injury [57].

Table 2.2: Glasgow Coma Scale [56].

Response	**Scale**	**Score (Points)**
Eye opening response	Eyes open spontaneously	4
	Eyes open to verbal command	3
	Eyes open to pain	2
	No eye opening	1
Verbal response	Oriented	5
	Confused conversation, but able to answer questions	4
	Inappropriate responses, but discernible words	3
	Incomprehensible sounds/speech	2
	No verbal response	1
Motor response	Obeys commands for movements	6
	Purposeful movements to painful stimuli	5
	Withdraws from pain	4
	Abnormal flexion	3
	Extensor response	2
	No motor response	1
Injury	Concussion	13 - 15
	Severe TBI	< 9

Not just GCS is used but also the measure of loss of consciousness and posttraumatic amnesia (PTA) durations [58]. PTA is defined as the interval from injury until the patient is oriented and able to recall newly formed memories [59]. A mild TBI can be considered if loss of consciousness is 30 minutes or PTA is 1 day; a severe TBI corresponds to a loss of consciousness longer than 24 hours and/or a PTA longer than 7 days [58].

A more complete tool that combines the aforementioned clinical indicators with neuroimaging abnormalities was developed to improve the reliability and accuracy of TBI severity assessment, the Mayo classification system. It classifies TBI into 3 categories: moderate-severe (definite) TBI, mild (probable) TBI, and symptomatic (possible) TBI [60].

In sports, after a TBI event and the player have been out of the field, in the medical office or department, there is a need for an emergent neuroimaging in order to exclude more severe structural causes of brain dysfunction and in some cases of persistent symptoms. It is usefull because TBI symptoms can be difficult to distinguish from other medical or psychiatric disorders that may have existed before the trauma [61], [62].

Conventional neuroimaging of the brain with computed tomography (CT) and magnetic resonance imaging (MRI) scans usually contributes a little to the TBI evaluation, although it does not detect microscopic axonal injury (injury in the transmission of information to different neurons, muscles, and glands). It can be done to anyone who has been suspected to have TBI but is really recommended for individuals with some symptoms or signs presents in the Table 2.3, [63].

Table 2.3: Warning signs from history and physical examination that may warrant neuroimaging and/or frequent reevaluations [63].

Loss of consciousness more than 30 s
Post traumatic amnesia more than 30 min
Seizure
Vomiting
Severe Headache
Pupillary asymmetry or other focal neurologic finding
Asymmetric or painful cervical range of motion

Neuroimaging modalities commonly considered in the evaluation of TBI include both structural and functional studies. Structural studies typically considered include CT and MRI. Advanced functional studies include functional MRI, magnetic resonance spectroscopy, positron emission tomography, single-photon emission CT, perfusion CT, and transcranial Doppler sonography. These studies may be of particular interest in the evaluation of TBI more severe when the initial structural studies are normal, although the evidence supporting the indications for their use is limited [64], [65], [66].

Despite the superior soft tissue contrast available by MRI, CT remains the first line imaging modality of choice due to its speed, global availability, lower cost, and infrequent contraindications precluding the need for screening procedures [67]. In addition, it has also been shown to provide important prognostic information, and with improvements in the quantitative evaluation. It may provide additional insights into the management and rehabilitation of symptoms post-TBI.

The next priority neuroimaging is the conventional MRI which is better able to identify abnormalities after TBI than CT alone. Global, regional, and more specific MRI abnormalities are related to TBI severity, making structural MRI a powerful diagnostic, prognostic, and scientific tool. Importantly, it is possible that MRI sensitivity and specificity could be dramatically improved by finding ways to combine the pathologic features from various imaging sequences [68]. These new imaging techniques are improving our understanding of TBI [69]. In the Figure 2.1 is presented the difference between a CT and a MRI in a 19-year-old male with severe closed head injury [70].

The goals of neurosurgical management in patients with TBI are to:

- Stop the haemorrhage,
- Remove the lesion causing mass effect,
- Relieve high intracranial pressure,
- Invasive intracranial monitoring device placement, if indicated.

Surgical removal of a portion of the skull, known as decompressive craniectomy, has been studied for the purpose of relieving increased intracranial pressure and improved outcomes in TBI.

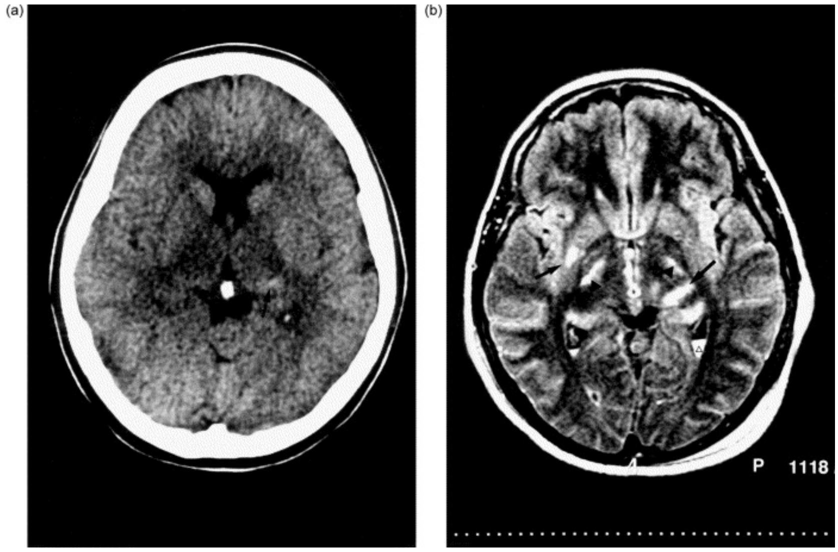


Figure 2.1: Difference between a CT (a) and a MRI (b) in a 19-year-old male with severe closed head injury [70].

In 2016, the RESCUEicp [71] trial compared decompressive surgery with medical management in patients with TBI and increased intracranial pressure. At 6 months, surgery was associated with lower mortality, higher vegetative state, higher lower severe disability, and higher upper severe disability rates, with similar rates of moderate disability and good recovery. The study concluded that decompressive craniectomy is a life-saving intervention for refractory intracranial hypertension in patients with TBI.

After all these years, TBI remains to be a complex event with more questions than answers. Instead of prohibiting the people to play sports, have been made some attempts to reduce the risk of TBI. It included protective gear improvements, game rule changes, trying to identify athletes at risk, and educate everyone involved with youth and high school sports about the dangers of TBI.

Protective gear: There is no good clinical evidence that currently protective equipment will prevent TBI. Biomechanical studies have shown a reduction in impact forces to the brain with the use of headgear and helmets. For skiing and snowboarding, studies suggest that helmets provide protection against head and facial injury and hence should be recommended for participants in alpine sports [72], [73], [74], [75]. In specific sports, such as cycling, motor and equestrian sports, protective helmets prevent other forms of head injury (skull fracture) that are related to falling on hard surfaces and may be an important injury prevention issue.

Rules changed: Several sports-governing bodies have implemented rule changes to eliminate or decrease specific contact that may place an individual at greater risk for suffering a TBI. An example of this is in soccer, where research studies demonstrated that upper limb-to-head contact in heading contests accounted for approximately 50% of concussions [76]. As noted earlier, rule changes also may be needed in some sports to allow an effective off-field medical assessment to occur without compromising the athlete's welfare, the flow of the game, or unduly penalizing the player's team. It is important to note that rule enforcement may be a critical aspect of modifying injury risk in these settings, and referees play an important role in this regard.

Education: Education and recognition remain the most important components of improving the care of athletes with TBI. It should target all the key individuals involved, including athletes, parents, coaches, school administrators, athletic directors, teachers, athletic trainers, physicians, and other health care providers. Previous studies have demonstrated poor knowledge of TBI recognition and management by players, coaches, and even physicians. They should be educated regarding the detection of concussion, its clinical features, assessment techniques, and principles of safe return to play. Methods to improve education including web based resources, educational videos, and international outreach programs are important in delivering the message. In addition, TBI working groups, plus the support and endorsement of enlightened sport groups, such as the Fédération Internationale de Football Association (FIFA), International Olympic Commission (IOC), International Rugby Board (IRB), and International Ice Hockey Federation (IIHF) that initiated this endeavor, have enormous value and must be pursued vigorously. Fair play and respect for opponents are ethical values that should be encouraged in all sports and sporting associations [21], [77], [78], [79], [80], [81], [82], [83], [84], [85], [86], [87], [88].

2.1.2 Concussion

"Concussion was recognized as a clinical entity for more than 1000 years when the term "concussion" has been used in confusing and contradictory ways [89]. It started to be more studied in boxers, but the interest of general population on this decrease because they accepted that the goal of the boxer is to inflict such an injury on their opponent. However, in 2002, after a postmortem evaluation of a retired National Football League (NFL) player, the possibility that concussion can result in a chronic brain injury and a progressive neurologic disorder raised." Since that, concussion has been a frequent topic of conversation in homes, schools, and on television. It has become a major focus of sports programs in communities and schools at all levels [90], [91].

"The definition of concussion changed over the years due to its complexity. Nowadays, we know that concussion is a form of mild traumatic brain injury (MTBI) that results when a direct blow to the head, neck, face or elsewhere on the body produces biomechanical forces that are subsequently transmitted to the brain [92]. That can also be an indirect loading on the brain with the inertia from whiplash or blast injury [93]. The resulting motion of the brain within the skull may result in stretching of and damage to axonal membranes and an associated cascade of detrimental effects. This cascade progresses and gives way to a period of depressed neuronal function in affected areas" [94].

"In the late 1990s, the American Academy of Neurology came out with their guidelines that suggested being unconscious was the only way to get you into the most severe grade of concussion. However, with its redefinition, now there is no need to be unconsciousness but just an alteration in the level of consciousness" [93]. With this fact, the number of symptoms related to concussion grew and the presence of one or more can be an alert signal [62]. Nevertheless, this signal can be related to something else as depression, anxiety and attention-deficit disorders due to the similarity of the signs and the symptoms [95]. Besides that, also the moment when the symptoms appear can change, it can occur or be recognized for several hours or days after the injury [96], [97].

In the Table 2.4, the range of symptoms related with a concussion was divided in four groups: physical, cognitive, emotional and sleep.

Table 2.4: Signs and symptoms of concussion [62].

Physical	**Cognitive**
Headache	Feeling mentally "foggy"
Nausea	Feeling slowed down
Vomiting	Difficulty concentrating
Balance problems	Difficulty remembering
Dizziness	Forgetful of recent information
Visual problems	Confused about recent events
Fatigue	Answers questions slowly
Sensitivity to light	Repeat questions
Sensitivity to noise	
Numbness\tingling	
Dazed	
Stunned	
Emotional	**Sleep**
Irritability	Drowsiness
Sadness	Sleeping more than usual
More emotional	Sleeping less than usual
Nervousness	Difficulty falling asleep
Anxiety	
Depression	
Personal Changes	

Inside these groups, the two most common symptoms are headache and dizziness. In addition, loss of consciousness only occurs in 10% of concussions which when prolonged, may indicate a need for neuroimaging to rule out the structural injury [13], [98].

In Table 2.5, there are some mandatory and discretionary signs of concussion defined by some sports, but for the same kind of sign, the action can be different. For example, in some sports like (Australia Football League, NFL, National Rugby League), mandatory signs indicate immediate removal from the field of play. In other sports like (National Hockey League), mandatory signs are indications for a mandatory off-field evaluation [99].

Table 2.5: Mandatory and discretionary signs of concussion in sports [99].

Mandatory signs of concussion	Discretionary signs of concussion
Loss of consciousness	Clutching the head
Lying motionless for more than 5s	Being slow to get up
Confusion\disorientation	Suspected facial fracture
Amnesia	Possible ataxia
Vacant look	Behavior changer
Motor incoordination	Other clinical suspicion
Tonic posturing	
Impact seizure	
Ataxia	

These two analyses show that recognize one concussion is a complex process that depends on two different factors: the impact and the person. The last one is the most influenced because every brain is different and the people can ignore the symptoms of concussion. One example of that is the football players who try to avoid the restriction from sports because they know if they show one single symptom, they can be removed from the game or the practice [92].

2.1.3 Chronic Traumatic Encephalopathy (CTE)

To talk about CTE we have to move to 1928 when Harrison Martland a coroner with interest in boxers, described the clinical pattern of cognitive, behavioral, and mood issues that these boxers manifested. He calls it "punch drunk", what years after would call Chronic Traumatic Encephalopathy by Cristchley in 1949 [100], [101]. In 2005 Omalu discover for the first time, CTE in 3 retired football players [90].

CTE is a neurodegeneration characterized by the abnormal accumulation of hyperphosphorylates tau protein (these proteins are found mostly in neurons and one of its main functions is to modulate the stability of axonal microtubules), within the brain. It only can be definitively diagnosed by postmortem examination of brain tissue [102]. This is a consequence of subconcussive hits that can appear in the 30's or 40's even when they stop to play. Roberts reported in 1990 that 17% of 224 retired boxers had CTE [103].

When a transient impact or acceleration-deceleration forces (mechanical or other type) are applied to the brain causing cellular, axonal, vascular, meningeal, or lymphatic impairment, it can result in a permanent damage or in a transient dysfunction of the neurovascular unit. Permanent damage can trigger cascades of pathologic events that produce neurologic marks that are detectable exclusively at the microscopic level. Such lesions may occur in any region like cerebrum, cerebellum, brainstem, spinal cord, cranial nerves [104]. In the Figure 2.2 there are some of the brain parts mentioned before.

However, new studies confirm that a history of repetitive head impacts could not result in the development of CTE. In this case, the individual differences in head trauma exposure are more influential in the risk of developing CTE. That happen because the individual head trauma characteristics like the severity of head trauma, the type of head trauma, the duration of repetitive head injury exposure, the total number of hits, the interval rest between hits are more influential in the risk of developing CTE [106].

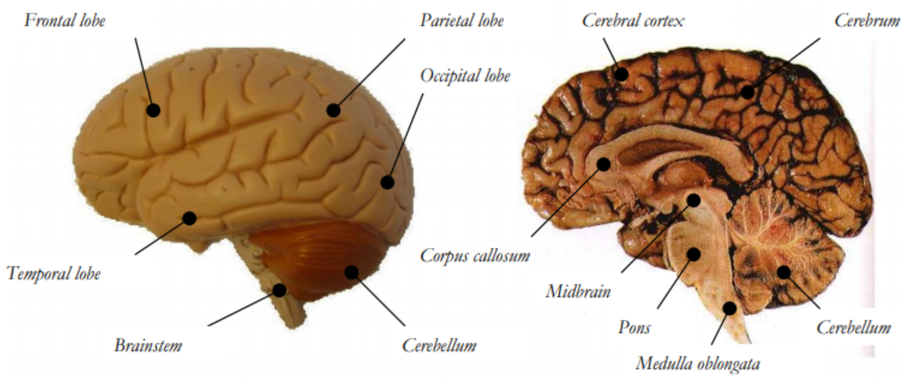


Figure 2.2: Brain anatomical divison [105]

The signs and symptoms of CTE involve at least 1 or 3 possible domains: cognition, behaviour/mood, and motor functioning. The cognitive difficulties seen in patients with CTE typically have a gradual progressive course and can include significant memory and concentration impairment, executive dysfunction, language difficulties, visuospatial difficulties (refers to a person's capacity to identify visual and spatial relationships among objects like understand the differences and similarities between objects), and motor disturbances. The onset of behavioural and cognitive symptoms is generally years after exposure to repetitive trauma and often presents in mid-life (after retirement from sports) [107], [108], [109], [110].

Furthermore, it was proposed 4 progressive stages of CTE and it was specified the primary neuropathologic features dependent on each stage of degeneration (Figure 2.3) [108]:

- In stage I, the brain has the normal weight and start to show some epicentre, around a blood vessel, of p-tau involving the sulcal depths especially of the superior and dorsolateral frontal cortices.

- In stage II, the brain continues with normal weight and more frequent epicentres at the depths of the sulci.

- In stage III, the weight of the brain reduced with mild cerebral atrophy and ventricular dilatation.

- Stage IV brains have a marked reduction in weight as a result of generalized cerebral cortical atrophy [101].

However, these 4 stages don't appear in all the cases of CTE like this [101].

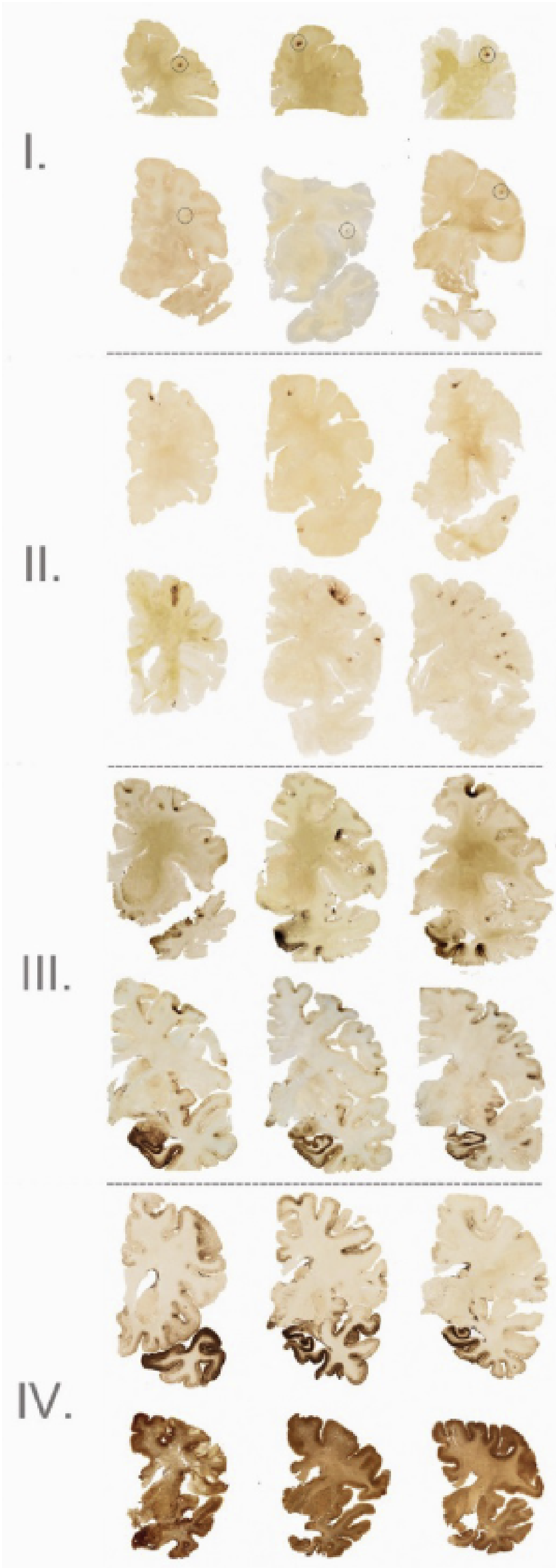


Figure 2.3: The distribution of pathologic tau accumulation in "4 stages" [108].

CTE is a postmortem neuropathologic diagnosis, there are no retrospective studies reviewing any structural or advanced neuroimaging biomarkers in patients diagnosed with the disease.

Structural MRI is the primary imaging modality for subacute to chronic TBI [111] due to the sensitive for detecting and characterizing brain injuries, particularly cerebral atrophy in chronic TBI. The number, size, and location of MRI abnormalities are correlated with the severity of TBI in the chronic stage and were used to predict clinical outcomes among patients with an early post-traumatic vegetative state [112].

Neuroimaging can help to identify CTE although, some neuroimaging studies show evidence of alterations in brain structure, function, and metabolism in football players, none showed a direct link to CTE [101].

After a diagnosis of probable CTE in a patient based on the combination of the clinical symptoms along with the MRI scan showing patchy areas of atrophy and hypometabolism in the cortex [113]. The next step is to look at the criteria for traumatic encephalopathy syndrome and its potential biomarkers presented in the Tables 2.6 - 2.9 and start the assessment process.

Table 2.6: General criteria for traumatic encephalopathy syndrome [114].

General criteria for traumatic encephalopathy syndrome: All five criteria must be met
1. History of multiple impacts to the head based upon the type of injury (a) and source of exposure (b)
a. Types of injuries:
i. Mild traumatic brain injuries or concussions, minimum of four
ii. Moderate/severe traumatic brain injury
iii. "Subconcussive" trauma
b. Source of exposures
i. Involvement of "high-exposure" contact sports for minimum of 6 years, including at least two at college level or higher
ii. Military service
iii. History of any other significant exposure to repetitive hits to the head
2. For moderate/severe traumatic brain injury, any activity resulting in the injury
3. No other neurological disorder present that likely accounts for all clinical features
4. Clinical features must be present for a minimum of 12 months
5. At least one core clinical feature must be present and considered a change from baseline
6. At least two supportive features must be present

Table 2.7: Core clinical features of traumatic encephalopathy syndrome [114].

Core clinical features of traumatic encephalopathy syndrome: At least one must be met
1. Cognitive. Difficulties in cognition as reported by either self or informant, by history, or clinician's report of decline and substantiated by impairment on standardized tests
2. Behavioural. Emotionally explosive, physically and/or verbally violent.
3. Mood. Feeling overly sad, depressed, and/or hopeless.

Table 2.8: Supportive features of traumatic encephalopathy syndrome [114].

Supportive features of traumatic encephalopathy syndrome: At least two must be present
1. Impulsivity. Impaired impulse control as demonstrated by new behaviours
2. Anxiety. History of anxious mood, agitation, excessive fears, or obsessive and/or compulsive behaviour
3. Apathy. Loss of interest in usual activities, loss of motivation and emotions, and/or reduction of voluntary, goal-directed behaviours
4. Paranoia. Delusional beliefs of suspicion, persecution, and/or unwarranted jealousy
5. Suicidality. History of suicidal thoughts or attempts
6. Headache. Significant and chronic headache, with at least one episode per month for 6 months
7. Motor signs. Tremor, rigidity, gait disturbance, falls, and/or other features of parkinson
8. Documented decline. Progressive decline in function and/or a progression in symptoms and/or signs, for a minimum of 1 year.
9. Delayed onset. Delayed onset of clinical features after significant head impact exposure, usually at least 2 years and in many cases several years after the period of maximal exposure

Table 2.9: Traumatic encephalopathy syndrome diagnostic subtypes [114].

Traumatic encephalopathy syndrome diagnostic subtypes: (1) Behavioural/Mood Variant, (2) Cognitive Variant, (3) Mixed Variant, (4) Dementia
Criteria for (4) Traumatic encephalopathy syndrome dementia:
1. Progressive course of cognitive core features, with or without behavioural and/or mood core features
2. Cognitive impairment (or cognitive impairment exacerbated by behavioural and/or mood) severe enough to interfere with the ability to function independently at work or in usual activities, including hobbies, and instrumental activities of daily living

The type of treatment in this type of disease depends on the specific signs and symptoms that the patient has. There are two possible treatments, pharmacological and non-pharmacological. The most complex is the first one because of the range of symptoms that exist. For example, a patient suspected to have CTE, perhaps they meet the criteria for traumatic encephalopathy syndrome, and they have a memory impairment as their major problem. In that case, cholinesterase inhibitor could be a possible choice to see if it helps.

Another example, a different patient, a younger one, and his main problem is depression or anxiety or behavioural problems, let's say he meets the criteria for the behavioural/mood variant of traumatic encephalopathy syndrome, an antidepressant medication could be possible option.

The second treatment focus on the brain exercises like aerobic that are very important not only because of the improvement cardiovascular fitness, but also there are studies that show it can actually increase the size of the hippocampus in young healthy individuals by increasing new brain cells. Besides that even walking just 30 min a day, 5 days a week, can be beneficial. All activities can help, like do crossword puzzles or play Sudoku, there is no doubt it is better to do these things than to watch TV. Social activities and interaction are also critically important, guarantee that the patients do not become isolated [113].

2.1.4 Subdural Haematoma (SDH)

A SDH results from bleeding within the subdural space caused by a rupture of an artery or bridging vein due to excessive loading, usually excessive rotation, which is the most common mechanism of SDH [115], [116], [117], [118]. Nearly one third of the acute SDH cases are directly related to bridging vein rupture [119]. Besides that, it is the most common major head injury and is associated with severe neurologic disability and death in many patients.

This type of injury arises from tangential force against the skull, and is directly related to rotational effects on the brain [120]. So, it can be considered a direct effect from inertial and non-contact forces. A SDH is caused by short duration and high strain rate loading [121].

"A subdural haematoma may occur as an isolated collection of blood within the subdural space. Many patients with complicated acute subdural haematoma sustain diffuse irreversible brain damage and do not improve after evacuation of the hematoma" [122].

"The clinical presentation of any patient, including an athlete, with acute subdural haematoma can vary and includes those who are awake and alert with no focal neurologic deficits, but typically patients with any sizeable acute subdural haematoma have a significant neurologic deficit "(Figure 2.4). This may consist of alteration of consciousness, often to a state of coma or major focal neurologic deficit.

One example of this is a football player with 2 mild concussions without loss of conciousness, separated by 7 and 10 days, sustained acute subdural haematoma [123].

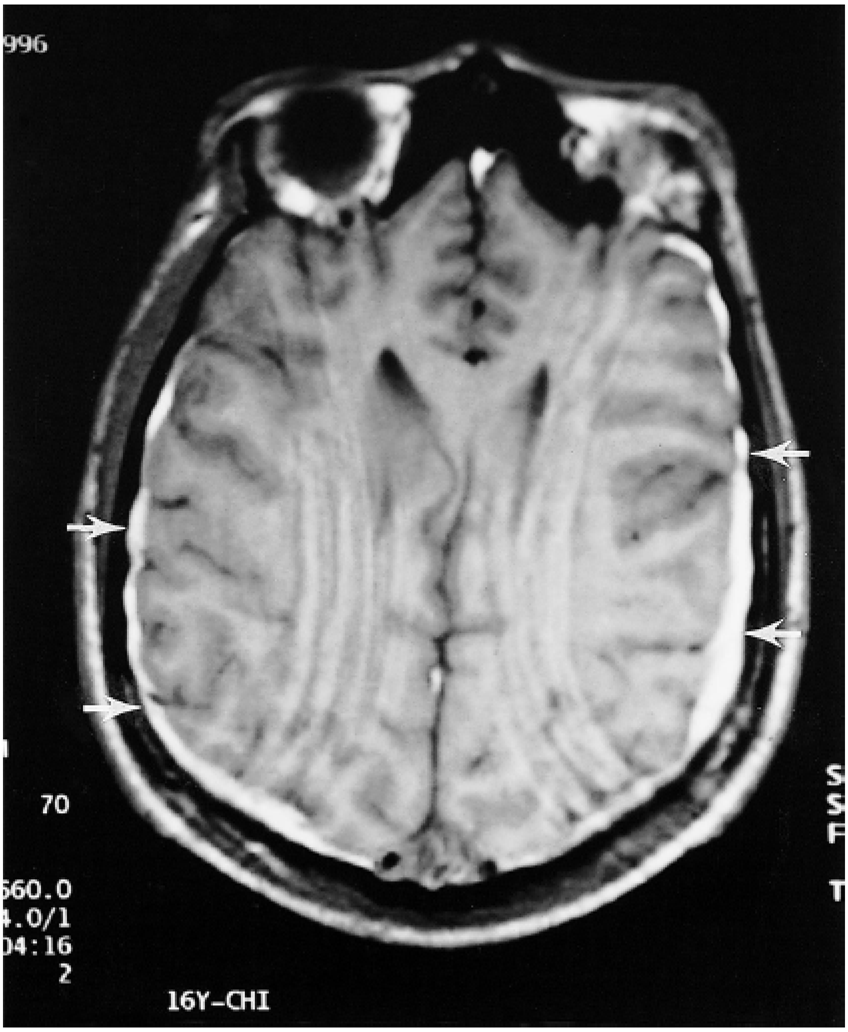


Figure 2.4: MRI of a high school football player with persistent headaches and poor school performance. Although the CT scan was normal, the MRI demonstrated small bilateral subdural haematoma, thus documenting the presence of an injury.

2.1.5 Diffuse Axonal Injury (DAI)

DAI is caused by the disruption or elongation of neuronal axons (responsible to transmit information to different neurons, muscles, and glands) in the brain tissue [124]. Axonal damage in DAI may be caused by the cascade of calcium-mediated events following stretching and pulling forces on the axonal membrane. These events result in obstruction of axonal transport and focal axonal swelling [94].

DAI arises from the same mechanisms than SDH, tangential forces applied to the skull. However, DAI is produced by a longer duration and more gradual onset of acceleration than SDH [105], [125].

It is one of the most frequent types of TBI [126]. More recently, [127] observed that this type of injury constitutes about more than 50% of all head injuries. Other authors consider DAI as the most common cause of persistent vegetative state and severe disability [128], [129], [130]. Thus, DAI is a frequent brain injury resulting from head impacts and often results in fatality or in long-term rehabilitation [120], [131].

MRI, especially the gradient echo and fluid attenuation inversion recovery sequences, are the imaging modalities of choice in cases of DAI with and without haemorrhage, respectively.

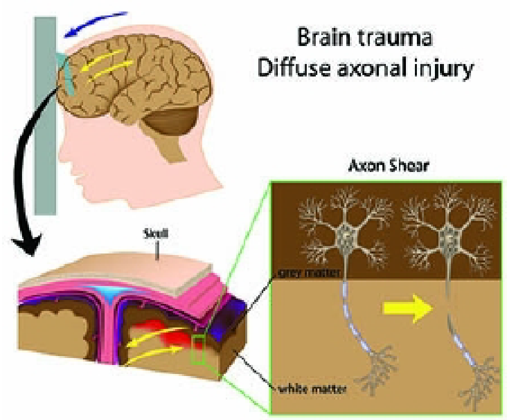


Figure 2.5: Diffuse axonal injury mechanism (adapted from Accident Attorneys [2016]).

Injury Measure Systems

For head injuries the scientists developed different criteria such as:
-Headform dependent:
a)Head Impact Power (HIP, rate of change of kinetic energy);
b)Gadd Severity Index (SI);
c)Head Injury Criterion (HIC);
d)Abbreviated Injury Scale (AIS);
e)Wayne State Tolerance Curve (WSTC);
f)Peak of Linear Acceleration (PLA);
g)Rotational acceleration threshold;
-Computational tools dependent:
h)Strain based injury criteria;
i)Brain von Mises stress;
j)Brain Pressure;

Every criterion has its own kinematic head injury assessment functions. With these results, the studies can be classified with magnitudes, the non-injury and concussive impacts, concussion thresholds, impact locations and position of the impacted player.

a) Head Impact Power (HIP)

With the big role of the linear and rotational kinetic energy in TBI events, could be a viable biomechanical assessment function for head injury using these two components (Equation 2.1). Each coefficient denotes the relative sensitivity for each of the six degrees of freedom of the head.

$$\begin{aligned} HIP =&Ma_x \int a_x dt + Ma_y \int a_y dt + Ma_z \int a_z dt + \\ &I_{xx}\alpha_x \int \alpha_x dt + I_{yy}\alpha_y \int \alpha_y dt + I_{zz}\alpha_z \int \alpha_z dt \end{aligned} \tag{2.1}$$

The first half represents the linear contribution and the second one the angular contribution. The coefficient M represents the mass of the human head and Ixx, Iyy and Izz represent the appropriate moments of inertia for it, which denote the injury sensitivity for each of the six head degrees of freedom. As HIP depend on the time, the thresholds is the maximum of the function [132]. In the Table 2.10 it is presented thresholds of this criterion for some injuries.

Table 2.10: HIP thresholds [132], [133].

Injury	**Threshold**
MTBI	50% probability: $\text{HIP}_{max} = 24$ kW
Severe TBI	50% probability: $\text{HIP}_{max} = 48$ kW
Concussion	50% probability: $\text{HIP}_{max} = 12.8$ kW
	95% probability: $\text{HIP}_{max} = 20.88$ kW

b) Gadd Severity Index (SI)

In 1966 was proposed a severity index to compare the severity of various head impacts. This index it's given by the equation 2.2:

$$GSI = \int a(t)^{2.5} dt \tag{2.2}$$

Where the a is the acceleration, force or pressure of the response function producing in g's and the t is the time in seconds. Its result can be used for two different objectives: comparing different head impact tests and if an impact exceeds the threshold of safety. In Gadd work, this threshold has the value of 1000 for serious internal head injury in frontal impact, in terms of g's [134].

c) Head Injury Criterion (HIC)

The HIC was proposed by the National Highway Traffic Safety Administration [135] as a criterion to identify the most damaging part of the acceleration by finding the maximum of the function with the equation 2.3 :

$$HIC = \left(\left[\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} a(t) dt \right]^{2.5} (t_2 - t_1) \right)_{max} \tag{2.3}$$

Where $a(t)$, is the resultant head acceleration in g's, the (t_2 - t_1) is the impact duration and t_1 and t_2 are the two points of the acceleration pulse, in time, in seconds. This approach only use two parameters for the definition of the injury onset, the acceleration and its duration over the time of impact.

HIC could be a useful predictor for comparing energy-absorbing of safety devices because it represents the global severity level of an impact and the potential head injury level [136], [137]. In the Table 2.11 it is presented some threshold established by this criterion.

Table 2.11: HIC thresholds

Injury	**Tolerance**	**Reference**
Head injury	31% probability of death: 2000	[138]
	65% probability of death: 4000	[138]
	99% Probability of life threatening injuries: 3000	[139]
	50% probability of AIS 3+: 1442	[140]
MTBI	95% Probability (for HIC_{15}): 485	[133]
	240	[141]
Severe neurological injury	50% Risk: 1032	[132]
Subdural Haematoma (SDH)	50% Risk: 1429	[132]
Concussion	200	[142]

d)Abbreviated Injury Scale (AIS)

The AIS was proposed in a effort to classify injuries according to its severity [143]. An AIS of 1 means a minor injury, while an AIS of 6 is attributed to lethal injuries. According to an updated AIS scale [144], Table 2.12 shows the meaning of each AIS code in terms of injury severity and type.

Table 2.12: AIS head injury classification

Code	**Injury Severity**	**Injury Description**
0	No Injury	
1	Minor Injury	Superficial laceration, nose fracture
2	Moderate Injury	Mandible fractures
3	Serious Injury	Basilar fracture, total scalp loss, single contusion cerebellum
4	Severe Injury	Brain damage: small Epidural haematoma (EDH) and SDH
5	Critical Injury	Penetrating injuries, brainstem compression, EDH, SDH, Difuse Axonal Injury (DAI)
6	Fatal Injury	Massive destruction of both craium and brain

e)Wayne State Tolerance Curve (WSTC)

The WSTC [144], was presented as failure criterion for prediction of skull fracture and concussion. This relation between cerebral concussion and skull fracture was also observed in [145], where 80% of all observed concussion cases also had linear skull fractures. The final form of the WSTC was developed by combining results from a wide variety of pulse shapes, obtained from cadavers, animals, human volunteers and clinical research, among others (Figure 2.6).

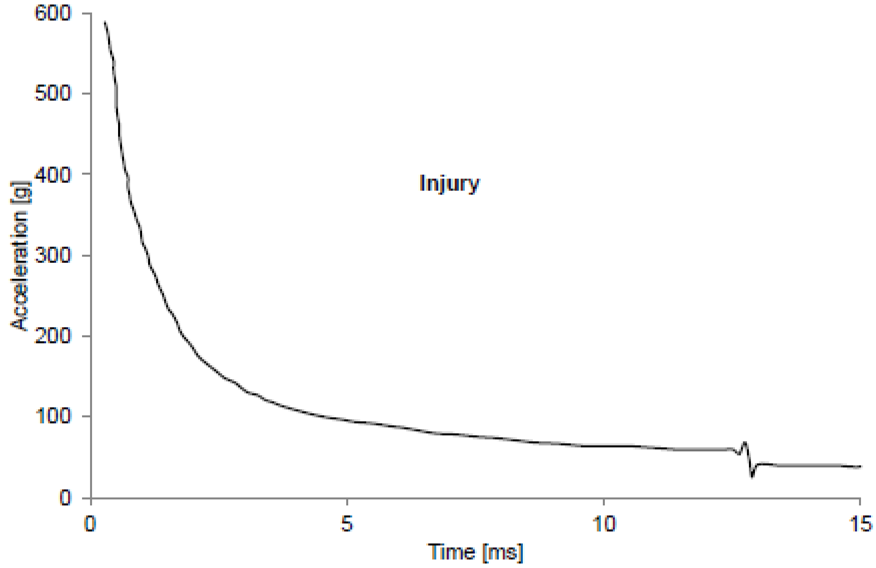


Figure 2.6: Wayne State Tolerance Curve (WSTC)

f)Peak of Linear Acceleration (PLA)

The PLA is the maximum value of the linear acceleration. This method ignores the impact duration, however, there are some studies that present the duration time for the peak of acceleration values. These prior values are presented in the Table 2.13 which shows the thresholds for this predictor.

Table 2.13: PLA thresholds

Injury	**Tolerance**	**Reference**
Head injury	a = 80 g for 3 ms	[146], [147]
	AIS 5: 250 - 300 g	[148]
	AIS 6: > 300 g	[148]
MTBI	95% Probability:	
	a = 1131 m/s^2	[133]
Concussion	a = 81 g	[142]
	a = 60.51 - 168.71 g	[149]
	a = 105 ± 27 g	[150]
	a = 74 ± 21 g	[151]
	50% Probability: a = 65.1 g	[152]
	75% Probability: a = 88.5 g	[152]

g)Rotational acceleration threshold

The relevance of rotational acceleration in brain injuries has been emphasized by many researchers, leading them to investigate thresholds to determine brain injuries. A study showed that angular acceleration must be applied long enough to attain a critical angular velocity and excessive displacement between brain and skull [153]. Some researchers showed that there are some thresholds values that combine velocity and acceleration to achieved to some brain injury (Table 2.14).

Table 2.14: Human brain tolerance to rotational acceleration and velocity

Injury	Threshold	Reference
Head Injury	50% Probability:	
	AIS 2+: w = 40 rad/s and $\alpha = 11.368$ rad/s^2	[140]
	AIS 3+: w = 55 rad/s and $\alpha = 18.775$ rad/s^2	[140]
MTBI	80% Probability: $\alpha = 7900$ rad/s^2	[141]
TBI	$\alpha = 1700$ rad/s^2 and w = 60 - 70 rad/s	[154]
	AIS 2: $\alpha > 1700$ rad/s^2 and w > 30 rad/s	[155]
	AIS 5: $\alpha > 4500$ rad/s^2 and w > 30 rad/s	[155]

h)Strain based injury criteria

There are some works where thresholds were proposed for the maximum principal strain (Table 2.15).

Table 2.15: Strain thresholds

Injury	Threshold	Reference
DAI	0.18	[156]
	50% probability:	
	0.21 (in the corpus callosum)	[41]
Concussion	50% probability:	
	0.15 (in the corpus callosum)	[157]

i)Brain von Mises stress

This criterion assumes that the von Mises stress is the cause of brain damage. Some of the proposed thresholds are given in Table 2.16.

Table 2.16: Stress thresholds

Injury	Stress [kPa]	Reference
MTBI	50% probability:	
	18	[158], [159]
Concussion	50% probability:	
	8.4 (in the corpus callosum)	[41]

j)Brain Pressure

This is a head injury predictor based on the intracranial pressure. Several studies were published with thresholds for this predictor. Some are presented in Table 2.17.

Table 2.17: Brain pressure thresholds

Injury	**Pressure [kPa]**	**Reference**
Moderate	172.3	[160]
Severe or fatal	234.4	[160]
Minor or absent	≤ 173	[161]
Severe	235	[162], [163]

In [164], using a FEHM, concluded that brain pressure has a better sensitivity for very short time impacts than the HIC. However, computed brain pressure does not correlate with some brain injuries. In addition, in [159] was established that computed brain pressure is not correlated with the occurrence of brain haemorrhages, whereas brain von Mises stress is.

2.2 Protective Devices

The increasing concern over head injury and the rapid growth of participants in contact-sport result in new and better protective devices to achieve to the biggest goal, prevent brain injuries. It could be near, but for now, reducing the impact energy is the best help that these devices could give.

2.2.1 Helmet

The first time that was used a football helmet made of leather was in 1893 during an Army-Navy game [165]. In 1980, was accepted the first standard for a helmet [166].

Nowadays all the sport helmets need to be certified to the National Operating Committee on Standards for Athletic Equipment (NOCSAE) standard.

Initial helmets were designed to reduce the risk of skull fractures [167], today it is more than that, its geometry and material are projected to offer the biggest head protection. These two characteristics are different between each other as it is presented in Figure 2.7 due to the range of game scenarios.

Figure 2.7: Types of sport helmets (Football, Lacrosse, Ice Hockey, Field Hockey, Cricket, Cycling, Baseball and MotoGP)

These are some examples of sport helmets that we can find nowadays. Most of them used materials like Vinyl Nitrile (VN) and Expanded Polypropylene (EPP). The big challenge today for the producers is to add more protection without compromising the performance of these devices.

2.2.2 Headgear

In some contact-sports, the use of helmet could be a reason for bad performance due to the size and the weight of it. So, there are some protective devices that can offer the same coverage than helmets and protection but more light and one of them is the headgear. As helmets, headgear could has different geometries (Figure 2.8):

Figure 2.8: Headgear types

In this type of type of devices, the Ethylene-Vinyl Acetate (EVA) foam is the most common material in the headgear pads. Following the Virginia Tech Helmet Lab was made the first independent rating to evaluate the performance of protective headgear for soccer players. The 22 models tested earned ratings ranging from two to five. The results demonstrated that some headgear were very effective: three models earned the top score of five stars, which translates to a reduction in concussion risk of at least 70 percent for the impacts tested [168]. Not only in soccer but also it is also used in football, lacrosse, rugby, water polo, volleyball, field hockey, etc.

2.2.3 Headbands

The second alternative to the helmets with less coverage than headgears are the headbands. At its market, there are some options with new geometries and different materials like we can see in the Figure 2.9:

Figure 2.9: Headbands types

In the Figure 2.9 the first headband in the left side had the highest score in the Virginia Tech Helmet Lab rating, which earned five stars, reduced injury risk by 84% for the impact test [169]. The material of these devices could change between different models but the material used in two of the figure's models are polyurethane (PU) and polyethylene (PE) foams. The others are made of polymers that give a good response to an impact. A headband can be used in the same sports as the headgear, which means, in sports that people mustn't use it, but they want to feel protected.

2.2.4 Protective device summary

In the Table 2.18 is presented a summary of this section that shows all the material that is used in each protective device.

Table 2.18: Protective devices table

Protective Device	**Material**
Helmet	Vinyl Nitrile (VN)
	Expanded Polypropylene (EPP)
Headgear	Ethylene-vinyl acetate (EVA) foam
Headband	Polyurethane (PU) foam
	Polyethylene (PE) foam

Chapter 3

Material Selection

This chapter describes the experimental tests performed to characterize agglomerated cork and the foams used in the headbands used as well as the simulations performed to validate the material models.

In this phase, find the best agglomerated cork for the headband application was the main goal. Thus, it was performed quasi-static and dynamic compression tests on 3 types of agglomerated cork with different densities, supplied by Amorim Cork Composites, a Portuguese company. Besides that, the company also financed 3 market headbands (Storelli, Force Field and Full90 headbands - Figure 3.1) to compare with the agglomerates mentioned before. In the Table 3.1, there are all the different densities belong to each type of material tested.

Figure 3.1: Headbands for test

Table 3.1: Sample densities

Sample	Density (kg/m^3)
Agglomerate Cork:	
NL10	140.00
NL20	193.53
NL25	234.12
Headband:	
Storelli	218.00
Full 90	309.00
Force Field	64.90

In each test, it was used 3 or more samples of the same material due to their recovery properties and test repeatability. Some synthetic foams and cork recovers almost totally its initial dimensions after compression although, there is always some damage, which value depends on the strains reached [48]. This is called viscoelasticity.

The quasi-static and dynamic tests are performed in order to obtain the mechanical properties required to characterize the agglomerated cork and the synthetic headbands materials.

With the first one was possible to know which agglomerate have the best quasi-static behaviour in terms of energy absorption. Only this material will be subjected to the dynamic test where it was evaluated its performance and compared against the synthetic foams.

Finally, all the data from the experimental tests reasonably matched the numerical simulation performed in Abaqus software.

3.1 Experimental Tests

All the experimental tests were carried at the department of mechanical engineering facilities of the University of Aveiro. The quasi-static test was made in the Shimadzu universal testing machine (Figure 3.2 in the left). The dynamic test was performed used a machine built by students called "drop tower" (Figure 3.2 in the right).

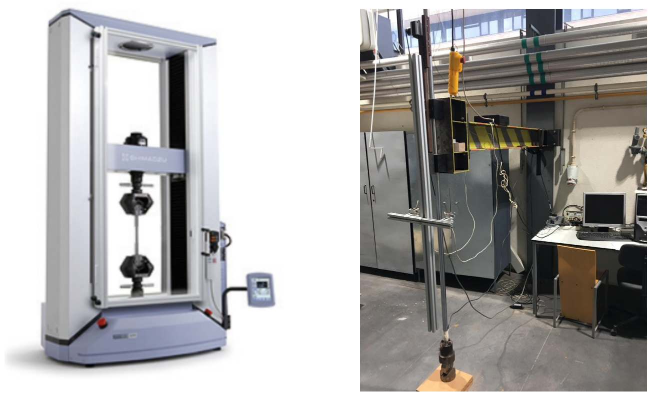


Figure 3.2: Quasi-Static and Dynamic test machines

3.1.1 Quasi-static tests

The quasi-static compression test did not follow any standard due to the samples' thickness and section. All the input in the machine have the objective of compress the maximum thickness of the samples. The table 3.2 shows some of the parameters took into account in the test.

Table 3.2: Quasi-static test parameters

Parameters	**Value**
Diameter of the movable plate	40 mm
Compression rate	2 mm/min
Total number of samples	30
Cork sample thickness	5 and 10 mm
Force Field headband thickness	11 mm
Full 90 headband thickness	11 mm
Storelli (front/back) headband thickness	10/15 mm

From each test at the velocity mentioned before, the machine gives three independent measurements: the time, the force (applied to compress the sample) and the position of the movable plate (in relation to the fixed one). The data referent to the force and the position was transformed into stress and nominal strain to obtain the behaviour curve of the material.

To obtain the uniaxial stress (σ) in the sample during the test, it was used the Equation 3.1 that relates the force (F) with the compressed area (A_c).

$$\sigma = \frac{F}{A_c} \tag{3.1}$$

$A_c = \pi \ \mathrm{r}^2 = 1256.64 \ \mathrm{mm}^2$
F [N]
σ [MPa]

To obtain the values for the nominal strain (ϵ) it was used the Equation 3.2 that relates the initial thickness (L_0) with the thickness on each time in the test (L).

$$\epsilon = \frac{L - L_0}{L_0} \tag{3.2}$$

After applied these two equations to the data from the machine, it was visualized all the stress-strain curves. For the materials used in this study, there is a specific stress-strain curve characterized by three regimes [170], as illustrated in the Figure 3.3:

- I - For very small strains, there is a linear elastic regime, which corresponds to cell edge bending. In Figure 3.3, σ_{el} represents the elastic buckling collapse stress;
- II - Within this range of strains, the compressive stress is almost constant during the compression. In Figure 3.3, it is possible to observe a stress plateau. This stress plateau corresponds to progressive cell collapse by elastic buckling;
- III - For such strains, cells are collapsed throughout the material and subsequent loading of the cell edges and faces against one another leads to high stresses.

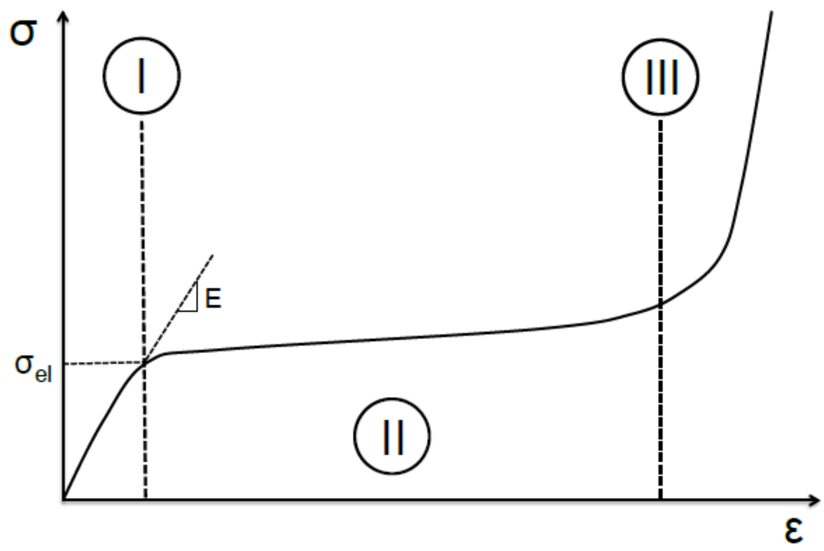


Figure 3.3: Typical uniaxial stress-strain curve of cellular material in compression [48].

The first and the second regimes were the crucial factors to select the best agglomerated cork and, compare it in terms of energy absorption with the market headbands (Figure 3.4).

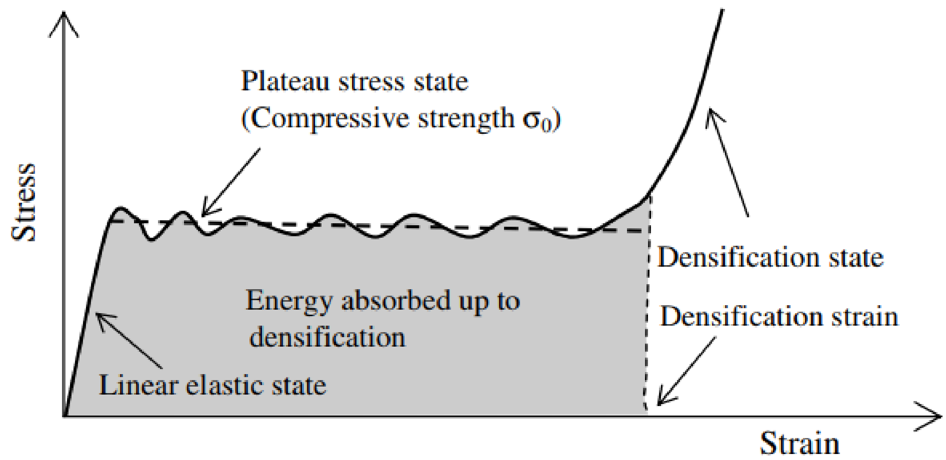


Figure 3.4: Measurement of energy absorbed with stress-strain curve [171].

Results

As the Figure 3.5 shows, the material curves are very different if we divide the agglomerate from the synthetic foam curves.

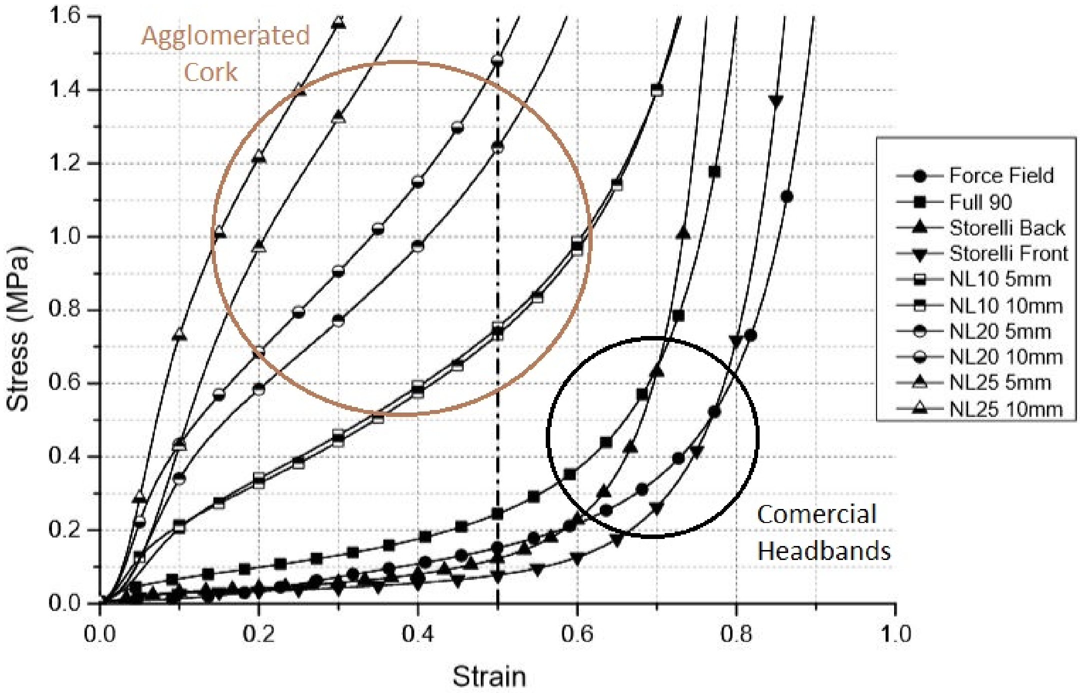


Figure 3.5: Measure of energy absorbed with stress-strain curve in quasi-static test.

All the results were analysed with the limit of 1.6 MPa (corresponds to a force of 2 kN), because it was a limit where all the materials start the densification phase and helps to compare the curves.

Setting a threshold of 50% strain, the first conclusion to take from this graphic was that cork can absorb more energy than the synthetic foams used in the market headbands, but at the expense of longer reaction forces. In terms of headband foams, the results were not so good as well because they have worse results than cork. Nevertheless, Full 90 was the best in terms of energy absorption in this test within the headband group. The results of the agglomerated cork showed that NL10 was the best agglomerate not just because the higher amount of energy absorption but also due to its density. The other agglomerates have higher densities and, at a certain level of impact they wouldn't be able to absorb but, it would transmit the impact force.

Thus, NL10 was the only agglomerated cork tested in the dynamic tests.

3.1.2 Dynamic tests

The dynamic test did not follow any standard due to the samples' thickness and section and the mass of the impactor. It was performed with the parameters that recreate the highest impact energy that the material was able to withstand. As was mentioned before, the test machine was built by students and it's divided into 5 principal systems: the impactor support beam, the lifting winch, the impactor and it mechanism of realese and the acquisition data system [172], (all presented in the Figure 3.6).

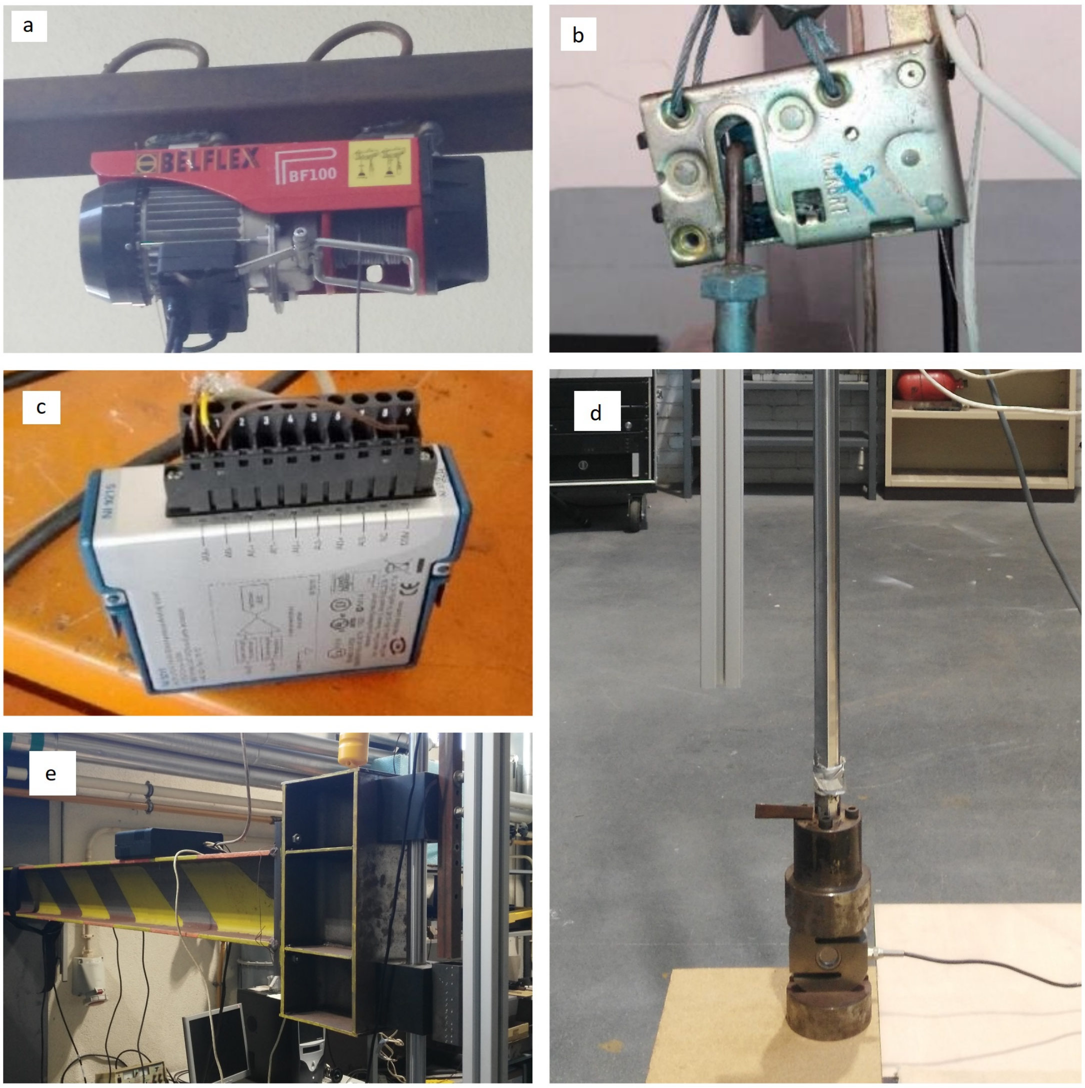


Figure 3.6: 5 systems of dynamic test machine (a - lifting winch, b - mechanism of release the impactor, c - acquisition data system, d - the impactor and e - the impactor support beam.)

The impactor is composed by an aluminum disk, which is in contact with the sample during the impact. However, in this study it was added a smaller extra disk in order to be compatible with the samples dimensions. Joined to the top of the disk there is a load cell that are responsible for the measurement of the impact force. One component made of steel is responsible to join the beam to the previous system. The total mass of the impactor system is 20±0.2 kg.

The specific system is presented in the Figure 3.7.

Figure 3.7: Impactor system

The position of the previous system is measured by an encoder on the top of the beam (Figure 3.8), [173].

Figure 3.8: Encoder system

In the impact tests, the impactor was raised to a certain height and it didn't change for all the samples. It was calculated to make sure that the peak force didn't exceed the maximum of 20 kN, the force limit measured by the load cell without endangering the equipment. The impactor was dropped and compressed the material, at the same time, the machine measures the force and the position of it. In the figure 3.9, there is a scheme of the configuration of the machine for the tests as was described before.

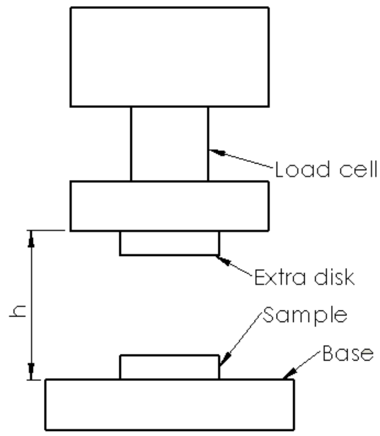


Figure 3.9: Impactor scheme

After all the tests were performed, the data from the machine had to be filtered and then transformed into stress and strain. However, this task wasn't so easy as expected, the rate of data acquisition is so high that the results had a lot of noise (Figure 3.10). The solution for this problem was a Butterworth filter that creates a curve more smooth and close to the material curve (figure 3.11).

The next step was to calculate the stress and the strain with the same equations used in the previous test.

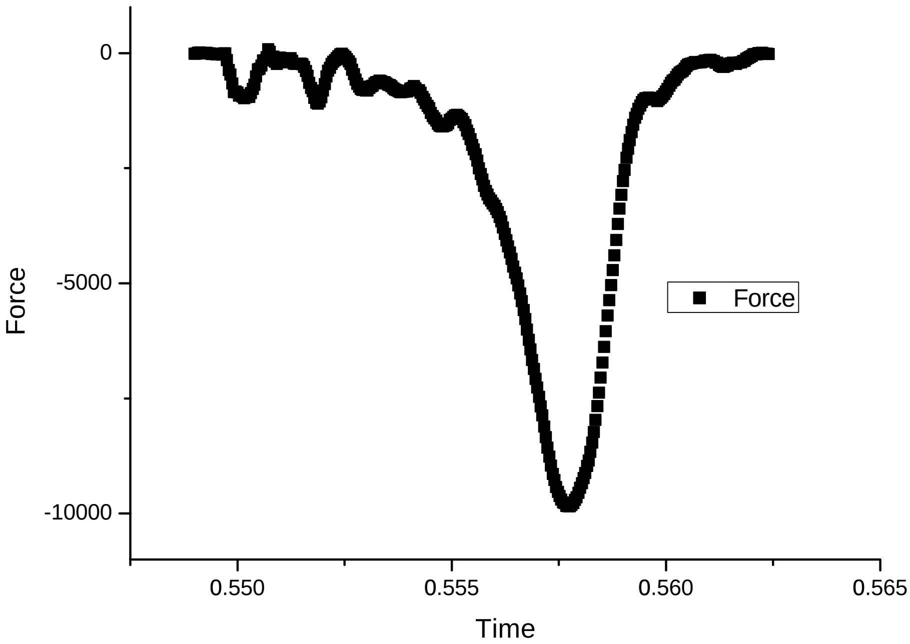


Figure 3.10: Noise from the machine data acquisition

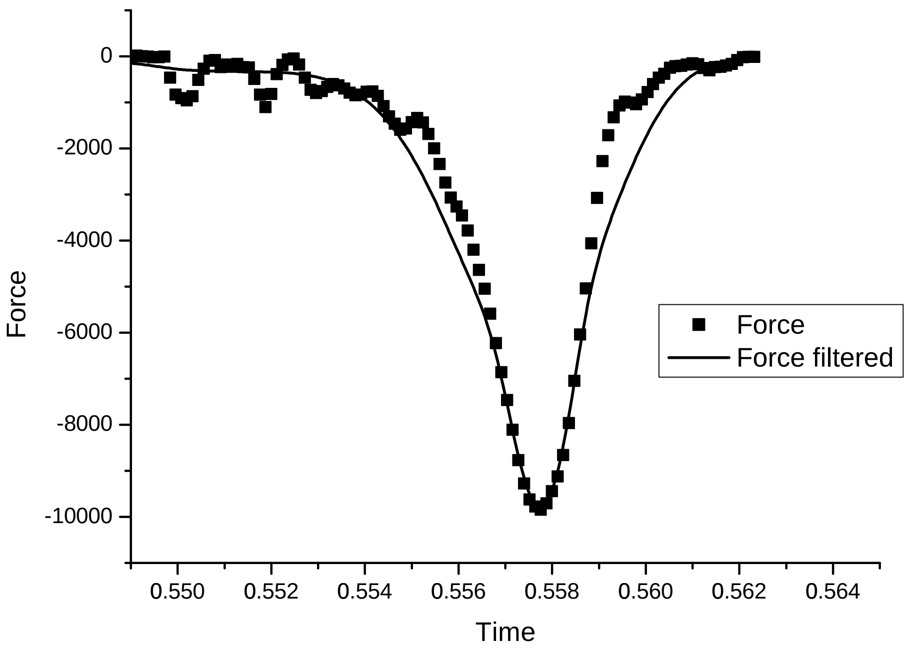


Figure 3.11: After butterworth filtration

Results

The final curves (Figure 3.12) that show the behaviour of the samples after a dynamic test were analysed the same way than in the previous test. Thus, the focus once again was in the area below the curve that represents the first and second stages.

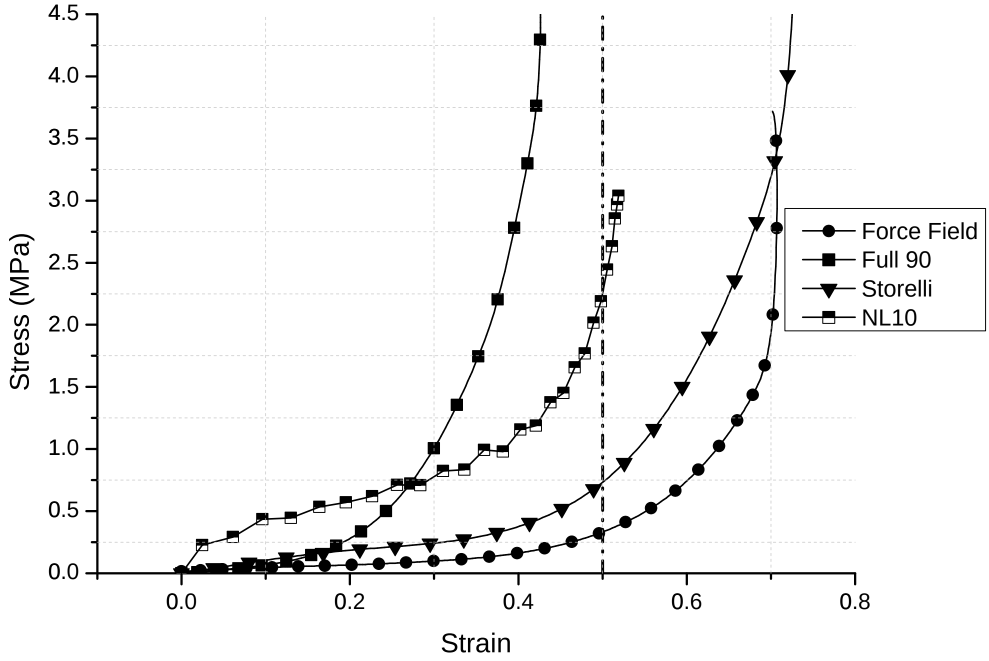


Figure 3.12: Results from Dynamic Test

As observed in the previous results, agglomerated cork can absorb more energy than the material from the market headbands. The Full 90 foam had an early densification due to its high density. Inside the group of the synthetic foams, Storelli was the best. In addition, NL10 showed a good potential in a dynamic situation.

3.1.3 Material Definition

To summarise the results from both experimental tests, the quasi-static and dynamic curves are presented together in the next four Figures 3.13 - 3.16. These data were crucial for the next steps, because with the material properties it was possible to validate both tests and to create a model of a headband made of each material.

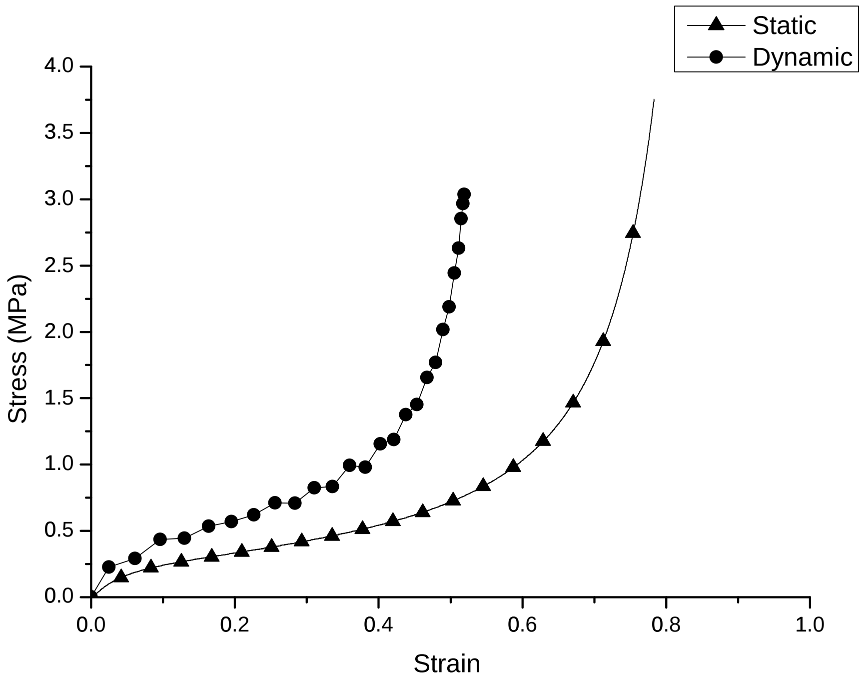


Figure 3.13: NL10 results from quasi-static and dynamic test

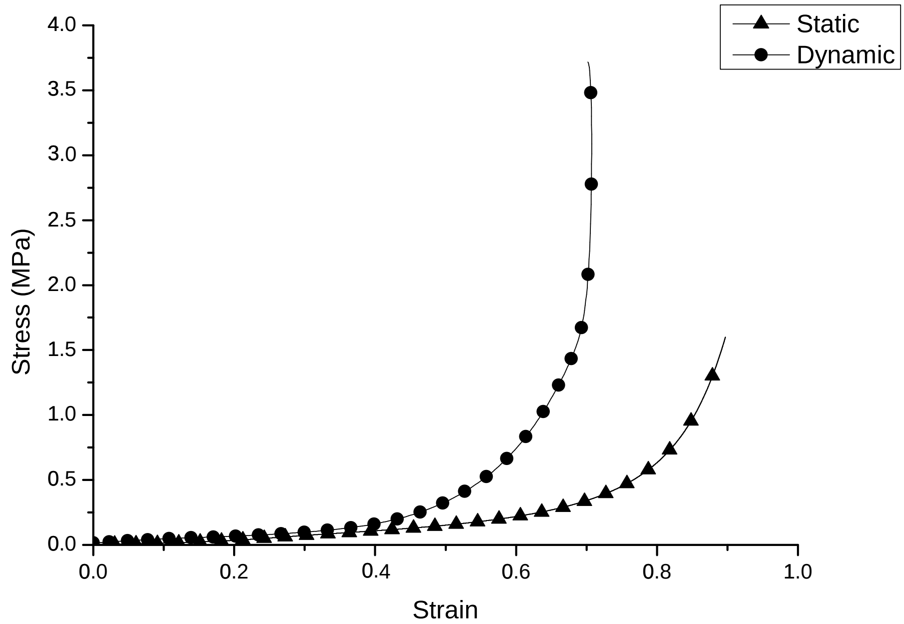


Figure 3.14: Force Field results from quasi-static and dynamic test

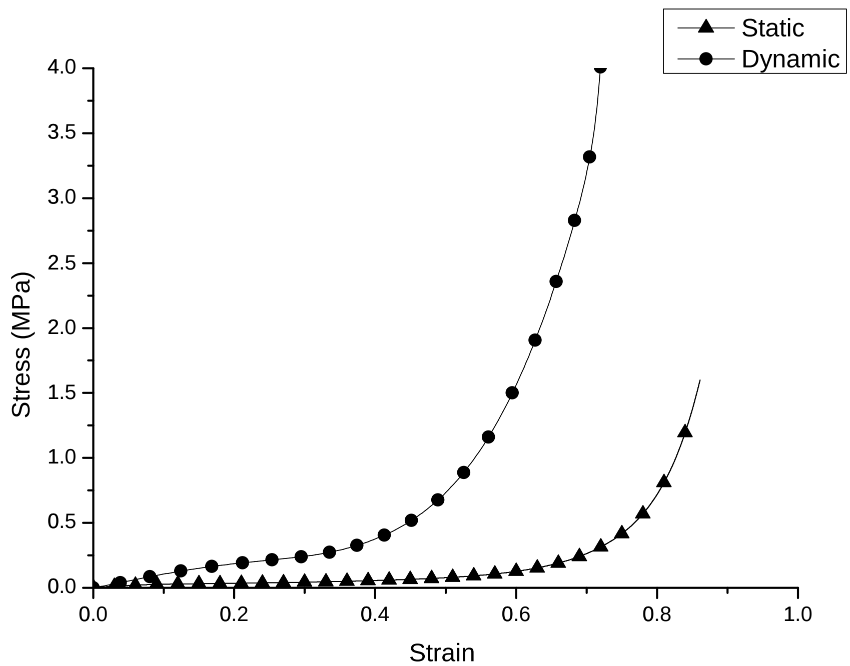


Figure 3.15: Storelli results from quasi-static and dynamic test

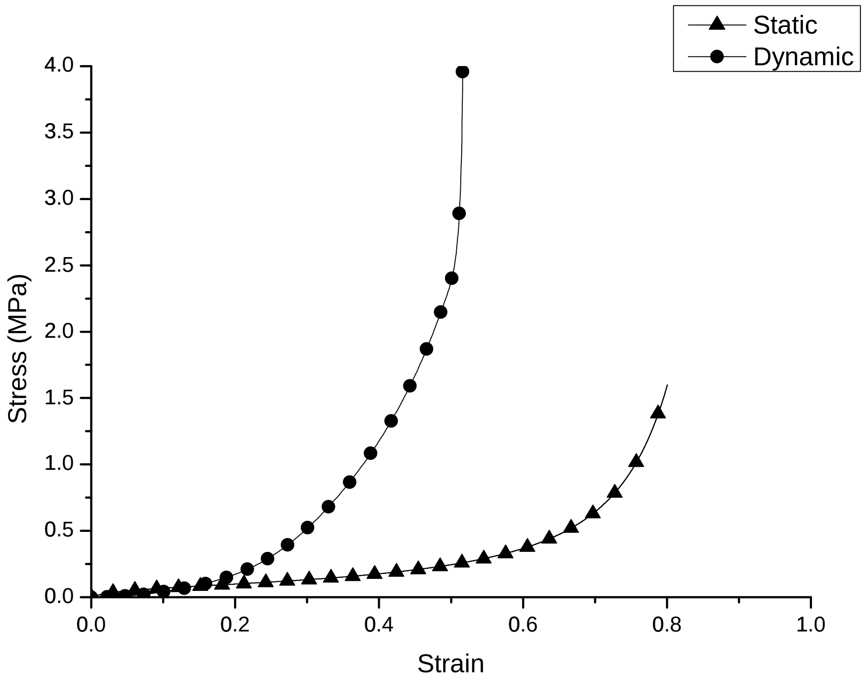


Figure 3.16: Full90 results from quasi-static and dynamic test

3.2 Numerical validation

The validation of the tests presented in the previous sections were made in the Abaqus software [174]. It has material models that can simulate the mechanical behaviour of a great variety of materials. For the synthetic foams and cork that were tested, hyperelastic and hyperfoam are the most common models used to simulate it.

The hyperelastic material model is isotropic and nonlinear, valid for materials that exhibit instantaneous elastic response up to large strains such as rubber, solid propellant, or other elastomeric materials, [174].

Hyperelastic materials are described in terms of a strain energy potential, which defines the strain energy stored in the material per unit of reference volume (volume in the initial configuration) as a function of the strain at that point in the material. In Abaqus, there are several forms of strain energy potentials available to model approximately incompressible isotropic elastomers: the Arruda-Boyce form, the Marlow form, the Mooney-Rivlin form, the Neo-Hookean form, the Ogden form, the polynomial form, the reduced polynomial form, the Yeoh form, and the Van der Waals form [174].

In these validations were used 3 different models; hyperfoam, reduced polynominal form and Ogden from.

3.2.1 Reduced Polynominal form

The form of the reduced polynomial strain energy potential is:

$$U = \sum_{i=1}^{N} C_{i0}(\overline{I}_1 - 3)^i + \sum_{i=1}^{N} \frac{1}{D_i}(J^{el} - 1)^{2i} \tag{3.3}$$

Where U is the strain energy per unit of reference volume; N is an integer (the polynomial order); C_{i0} and D_i are temperature-dependent parameters; $\overline{I}_1$ is the first deviatoric strain invariant defined as:

$$\overline{I}_1 = \overline{\lambda_1}^2 + \overline{\lambda_2}^2 + \overline{\lambda_3}^2 \tag{3.4}$$

where the deviatoric stretches $\overline{\lambda_i} = J^{-\frac{1}{3}}\lambda_i$; J is the total volume ratio; J^{el} is the elastic volume ratio and λ_i are the principal stretches. The initial shear modulus and bulk modulus are given by:

$$\mu_0 = 2C_{10}, \qquad K_0 = \frac{2}{D_1} \tag{3.5}$$

3.2.2 Ogden form

The form of the Ogden strain energy potential is:

$$U = \sum_{i=1}^{N} \frac{2\mu_i}{\alpha_i^2}(\overline{\lambda_1}^{\alpha_i} + \overline{\lambda_2}^{\alpha_i} + \overline{\lambda_3}^{\alpha_i} - 3) + \sum_{i=1}^{N} \frac{1}{D_i}(J^{el} - 1)^{2i} \tag{3.6}$$

where $\overline{\lambda}_i$ are the deviatoric principal stretches $\overline{\lambda}_i = J^{-\frac{1}{3}}\lambda_i$; λ_i are the principal stretches; N is an integer (the polynomial order); and μ_i, α_i, D_i and are temperature-dependent material parameters. The initial shear modulus and bulk modulus for the Ogden form are given by

$$\mu_0 = \sum_{i=1}^{N} \mu_i, \qquad K_0 = \frac{2}{D_1} \tag{3.7}$$

3.2.3 Hyperfoam

Hyperfoam is an isotropic and nonlinear material model typically used to characterize elastomeric foams that present hyperelastic behavior. Besides that it is used in finite-strain applications where it can deform elastically to large strains, up to 90% strain in compression. It is defined by a strain energy potential, also known as strain energy density function, which defines the strain energy stored in the material per unit of reference volume (initial volume) as function of the strain in the material [35]. In the hyperfoam material model, the elastic behaviour of the foams is based on the following strain energy function or potential:

$$U = \sum_{i=1}^{N} \frac{2\mu_i}{\alpha_i^2}\left[\hat{\lambda}_1^{\alpha_i} + \hat{\lambda}_2^{\alpha_i} + \hat{\lambda}_3^{\alpha_i} - 3 + \frac{1}{\beta_i}((J)^{-\alpha_i\beta_i} - 1)\right] \tag{3.8}$$

where N is an integer (the polynomial order); μ_i, α_i, and β_i are temperature-dependent material parameters;

$$\hat{\lambda}_i = \left(J^{th}\right)^{-\frac{1}{3}} \lambda_i \longrightarrow \hat{\lambda}_1 \hat{\lambda}_2 \hat{\lambda}_3 = J^{el}; \tag{3.9}$$

and λ_i are the principal stretches. The elastic and thermal volume ratios, J^{el} and J^{th}, are defined below. The coefficients μ_i are related to the initial shear modulus, μ_0, by:

$$\mu_0 = \sum_{i=1}^{N} \mu_i \tag{3.10}$$

while the initial bulk modulus, K_0, follows from:

$$K_0 = \sum_{i=1}^{N} 2\mu_i \left(\frac{1}{3} + \beta_i\right) \tag{3.11}$$

For each term in the energy function, the coefficient β_i determines the degree of compressibility. β_i is related to the Poisson's ratio, ν_i, by the expressions:

$$\beta_i = \frac{\nu_i}{1 - 2\nu_i}, \qquad \nu_i = \frac{\beta_i}{1 + 2\beta_i} \tag{3.12}$$

Thus, if β_i is the same for all terms,we have a single effective Poisson's ratio, ν. This effective Poisson's ratio is valid for finite values of the logarithmic principal strains ϵ_1, ϵ_2, ϵ_3; in uniaxial tension $\epsilon_1 = \epsilon_2 = -\nu\epsilon_1$.

3.2.4 Model

To validate the experimental test it was created a model that represents each component present in both test machines: One deformable solid that represents the material sample and two discrete rigid shells that represent the impactor and the base or the ground (Figure 3.17). The elements used in the deformable body were hexahedrons with reduced integration (C3D8R). This main model was used in all the validations related to the static and dynamic test materials.

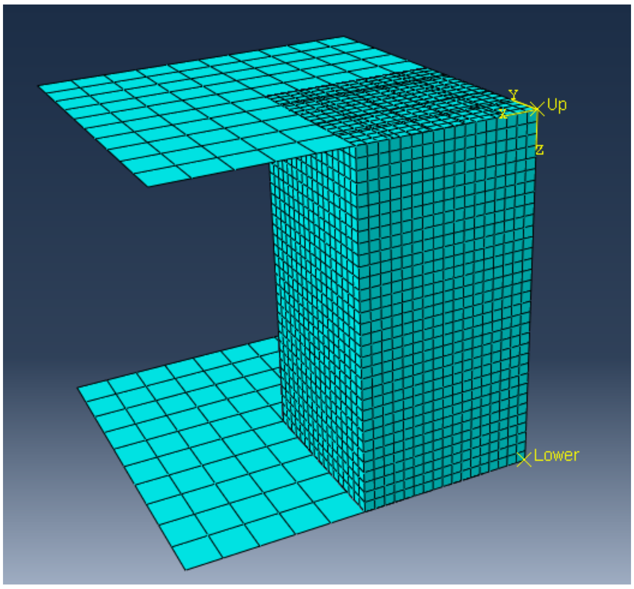


Figure 3.17: Model validation for static and dynamic tests

3.2.5 Quasi-static test validation

In the Table 3.3 the best formulation and the Poisson's ratio to fit the curve of the material behaviour in the quasi-static test is presented for each material:

Table 3.3: Best formulation model (quasi-static test)

Material	**Model**	**Strain energy potential order**	**Poisson's ratio**
Cork NL10	Hyperfoam	1	0
Cork NL20	Hyperfoam	1	0
Force Field	Hyperfoam	2	0
Storelli	Hyperfoam	3	0
Full90	Hyperfoam	3	0

With all the previous parameters and the quasi-static experimental curves it was possible to validate all the test as is presented in the next 5 (Figures 3.18 - 3.22):

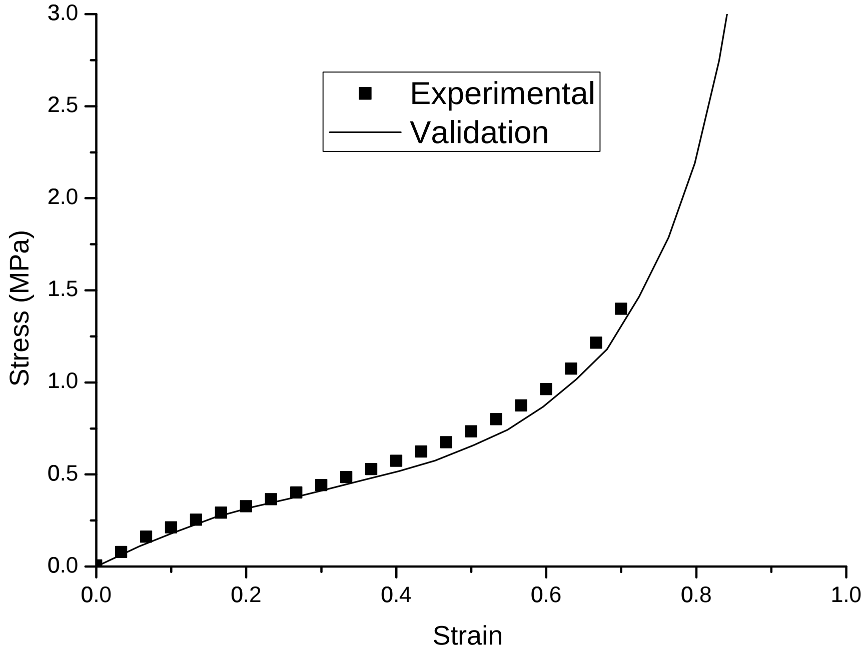


Figure 3.18: Agglomerate NL10 static test validation

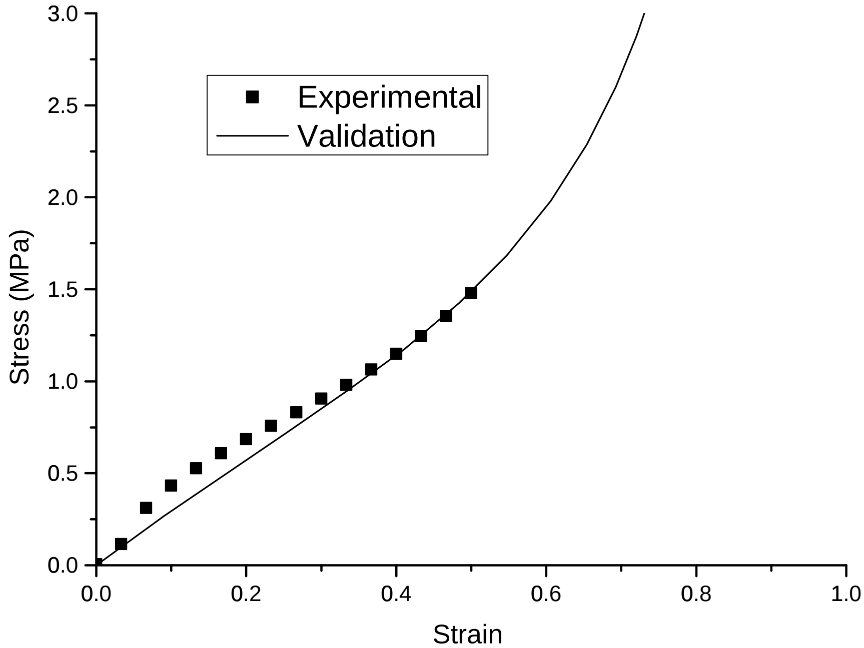


Figure 3.19: Agglomerate NL20 static test validation

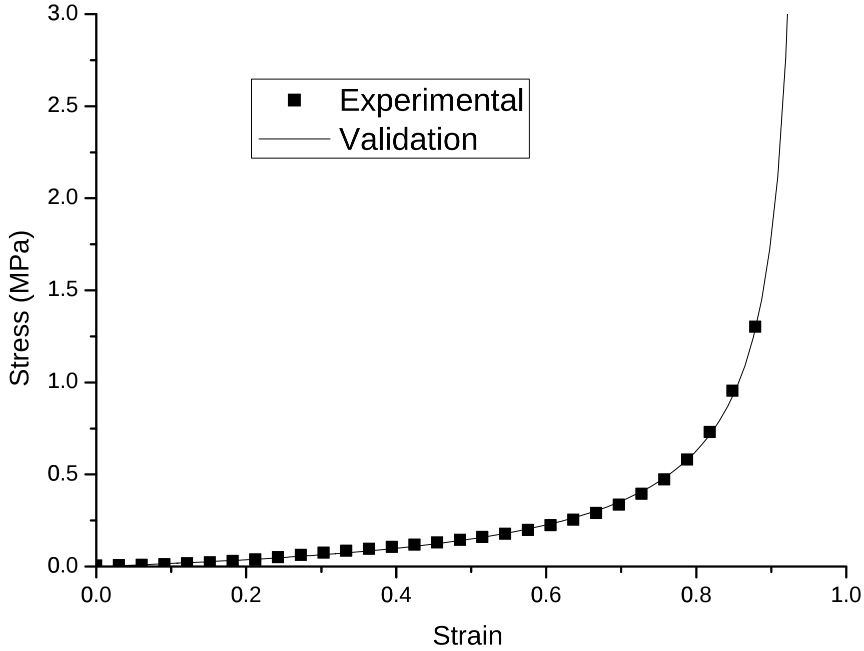


Figure 3.20: Force Field static test validation

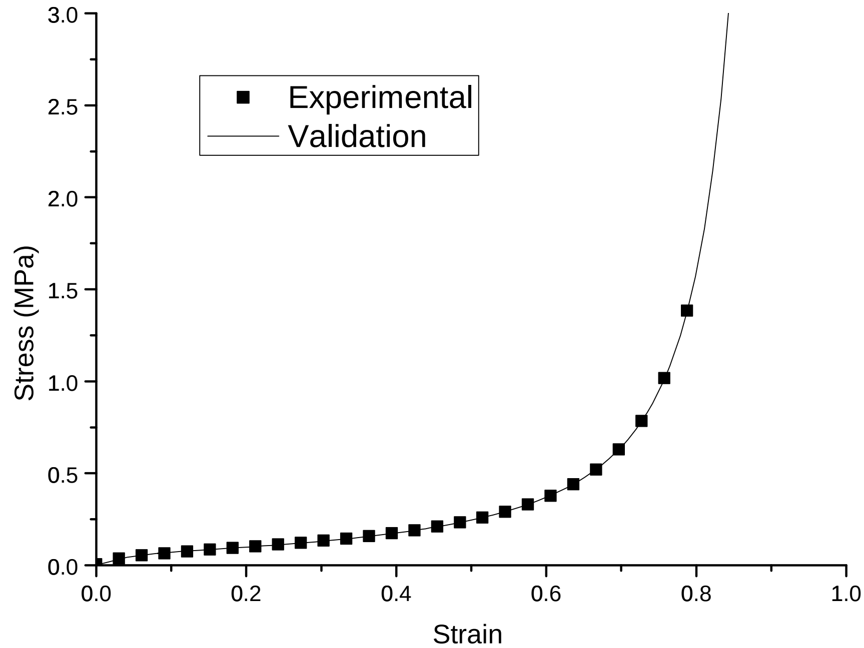


Figure 3.21: Full90 static test validation

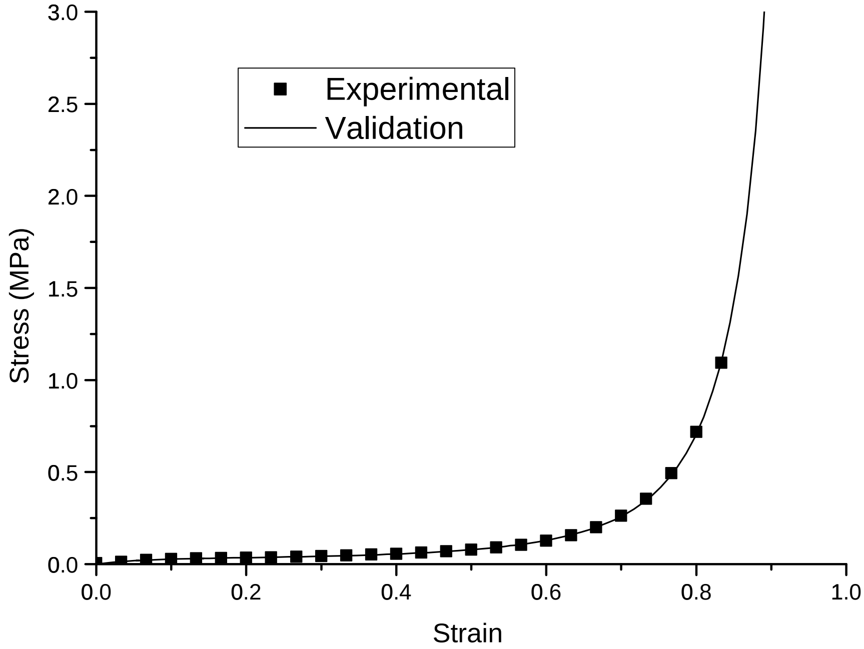


Figure 3.22: Storelli static test validation

All the quasi-static tests were validated. Only the validation curves of the agglomerated cork had an interval where the curve didn't fit exactly with the experimental curve although, the error isn't worrisome.

3.2.6 Dynamic test validation

In the Table 3.4 the best formulation and the Poisson's ratio to fit the curve of the material behaviour in the dynamic test is presented for each material:

Table 3.4: Best formulation model (dynamic test)

Material	**Model**	**Strain energy potential order**	**Poisson's ratio**
Cork NL10	Hyperfoam	3	0
Force Field	Ogden	2	0
Storelli	Reduced Polynominal	6	0
Full90	Reduced Polynominal	5	0

Once again, it was possible to validate all the test with all the previous parameters and the dynamic experimental curves as is presented in the Figures 3.23 - 3.26:

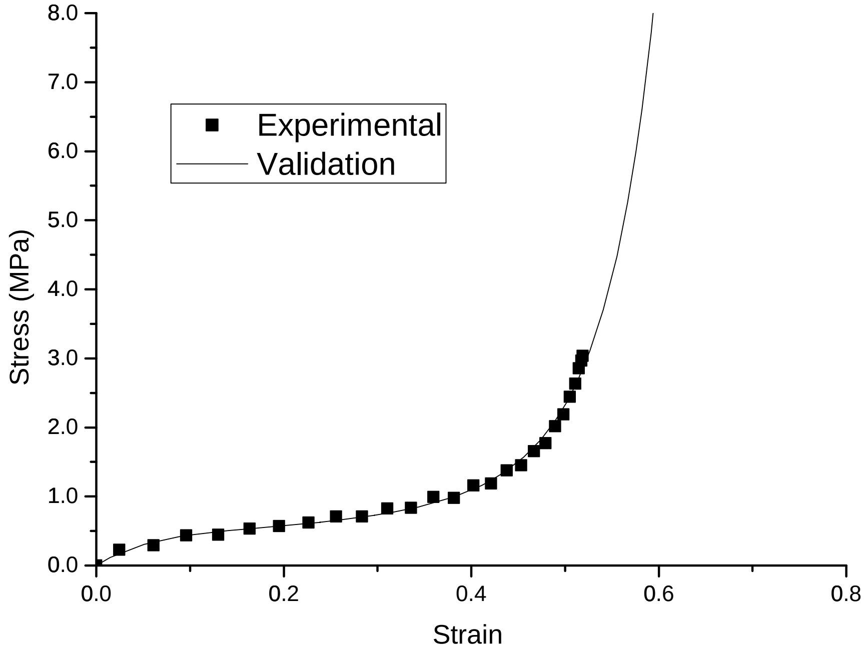


Figure 3.23: Agglomerate NL10 dynamic test validation.

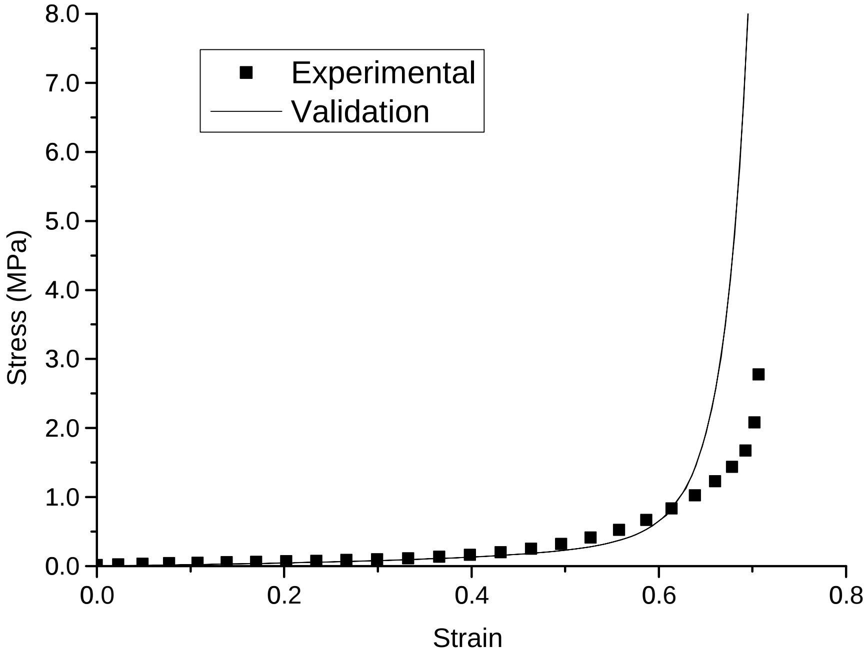


Figure 3.24: Force Field dynamic test validation.

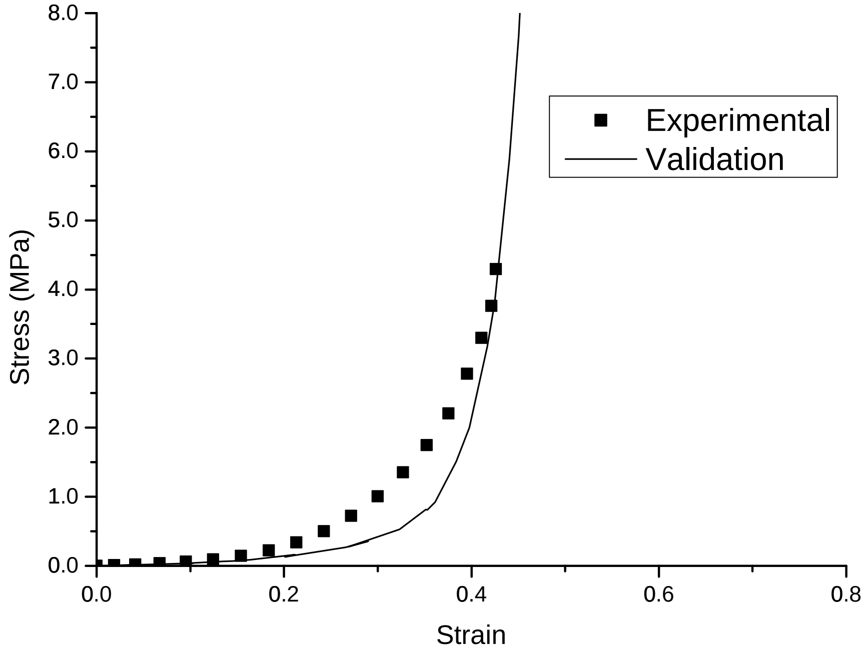


Figure 3.25: Full90 dynamic test validation.

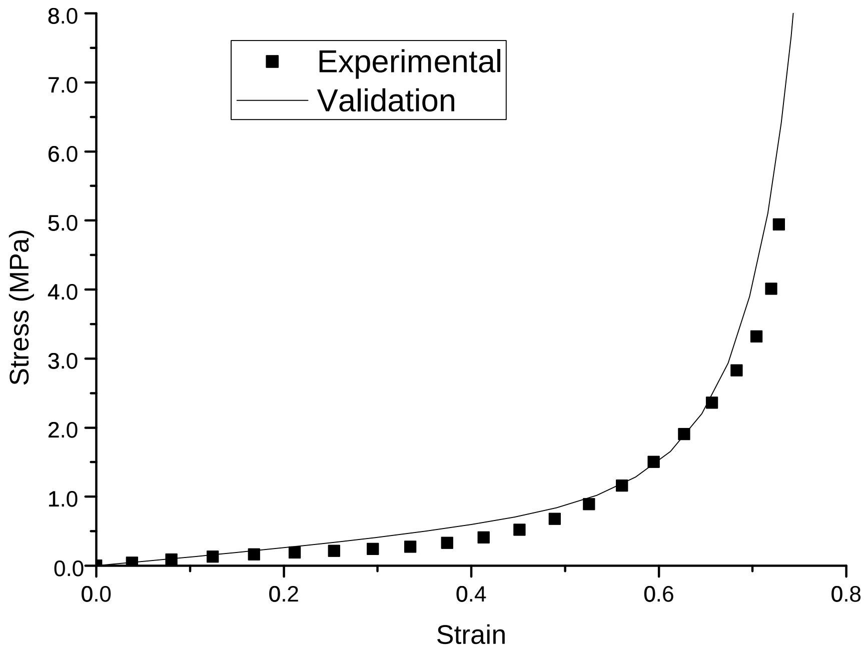


Figure 3.26: Storelli dynamic test validation.

The validation of the dynamic tests was reasonably good and all the tests were validated. However, the Full 90 and Force Field validation curve had a bigger error than expected because the model that most fitted with the experimental data start the phase of densification latter than the experimental. Thus, the formulations were changed for reduced polynomial and Ogden that start the densification in the correct value of strain despite the loss of some energy absorption.

This validation curves moved to the next step to define the material of the headband model.

Chapter 4

New Cork-based Headband

This chapter presents the development of a new headband model and its assessment based on the head injury risk predictions by YEAHM. This evaluation is performed in order to verify if a headband composed of agglomerated cork liners is an alternative to the ones in the market.

On this chapter is presented the new headband model. It was the base for all the tests, to see the potential of the cork in these type of devices. With 50 mm of height and 10 mm thick (Figure 4.1), the headband was simulated with all the materials tested before, under different conditions, in this case, impact energy values.

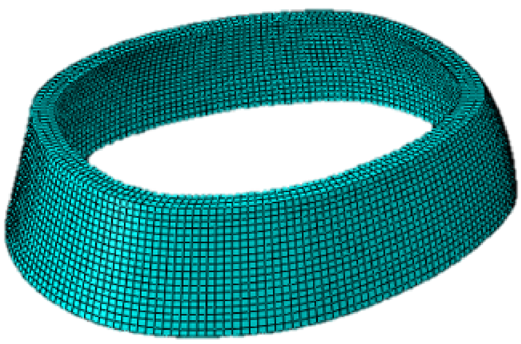


Figure 4.1: Headband model.

4.1 Headform tests

The first test made was an impact test with 7 different energy values (Table 4.3), that recreate several sports impacts and help to analyse the materials' behaviour under lower and high impact energies. In the Figure 4.2 it is presented the test scheme where the system headform-headband strike the ground or a wall. The objective of this simulation was to know the linear acceleration in the headform center of mass.

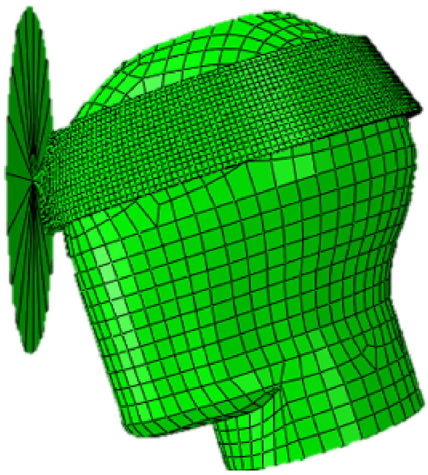


Figure 4.2: Impact test scheme.

The Table 4.1 shows the parameters and all the parts used in the simulation to recreate the impact.

Table 4.1: Parameters of the headform impact test simulation

Part	Type of Part	Element
Headband	Deformable	C3D8R
Headform	Discrete Rigid	R3D4, R3D3
Ground/Wall	Analytic Rigid	-

Another important parameter in this test was the contact algorithm. In this case it was used the kinematic instead of the penalty with a friction coefficient of 0.5. The different between this two methods is that the first uses a kinematic predictor/corrector contact algorithm to strictly enforce contact constraints (for example, no penetrations are allowed). The second has a weaker enforcement of contact constraints but allows for treatment of more general types of contact [174].

4.1.1 Headform

The headform used in the test was a ECE 22.05 headform model (Figure 4.3), [46], with 5.6 kg and the inertial moments presented in the Table 4.2 [175].

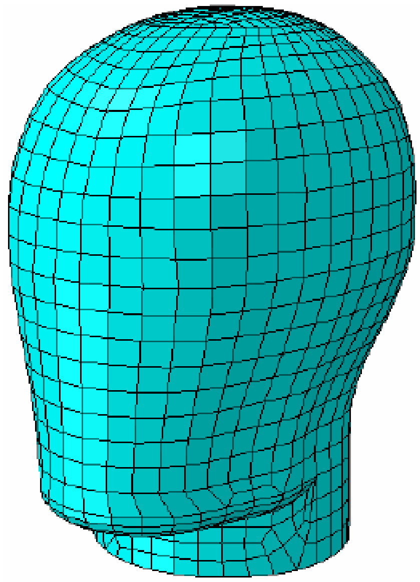


Figure 4.3: Headform.

Table 4.2: Moments of inertia of the headform

	I_{xx} [kg.cm^2]	I_{yy} [kg.cm^2]	I_{zz} [kg.cm^2]
Moments of inertia	286	338	209

4.1.2 Impact Energy

As was mentioned before, 7 values of impact energy were considered to make a scale of real situations. Assuming the mass of the headform 5.6 kg it was calculated the velocity necessary to achieve to each impact energy level. In the table 4.3 it is presented all the values and with its correspond velocity.

Table 4.3: Impact energy and velocity values

Impact Energy (J)	Velocity (m/s)
20	2.67
40	3.78
80	5.35
120	6.55
160	7.56
200	8.45
240	9.26

In each interval of impact energy was consider several situations like standards related to helmet and headgears as well as real sport impacts like heading a ball in a soccer game (Figure 4.4).

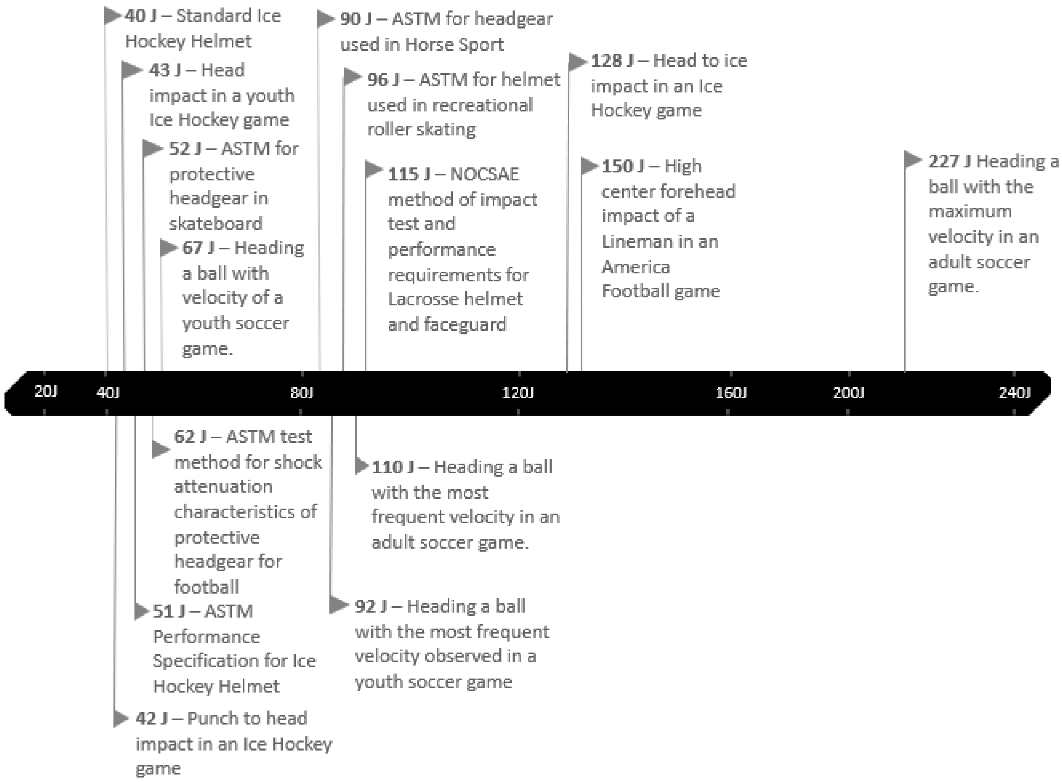


Figure 4.4: Impact energy in different situations [176], [177], [178], [179], [180].

4.1.3 Results

As it was expected the headform center of mass acceleration results were proportional to the energy impact. In the next figures it is presented the curves of all impacts for each material and impact energy.

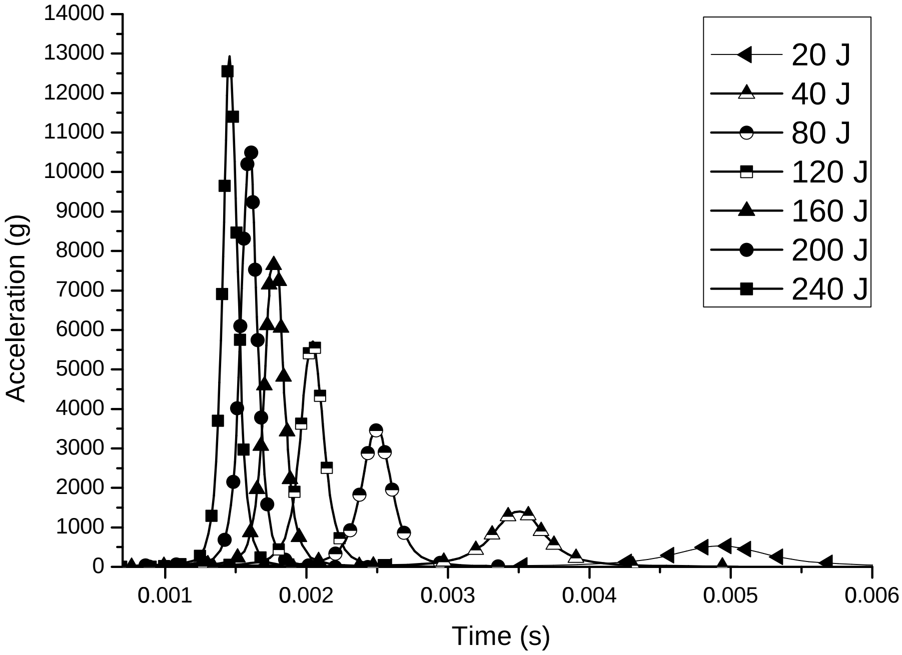


Figure 4.5: Headform impact test result for Cork

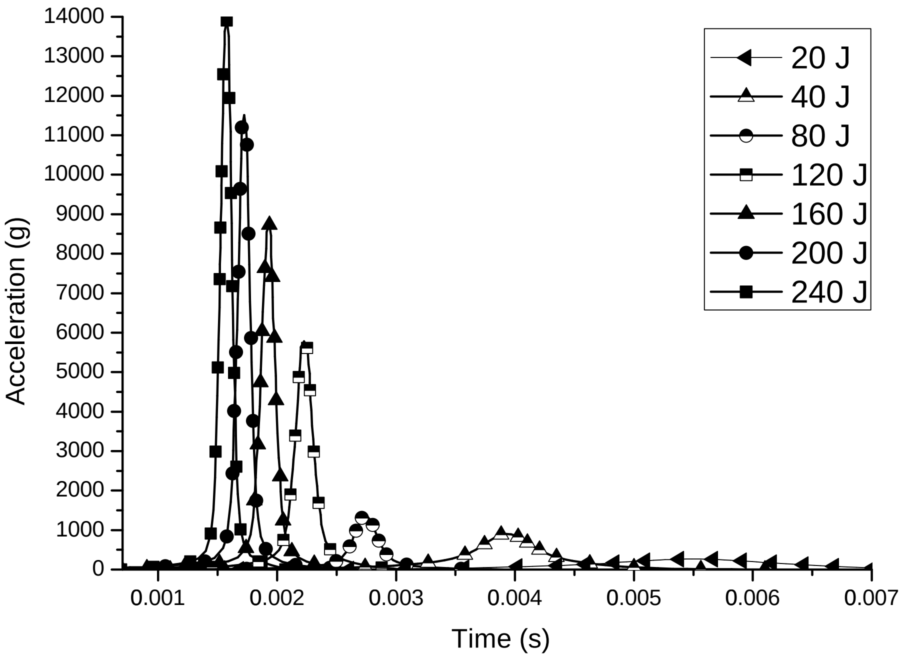


Figure 4.6: Headform impact test result for Storelli

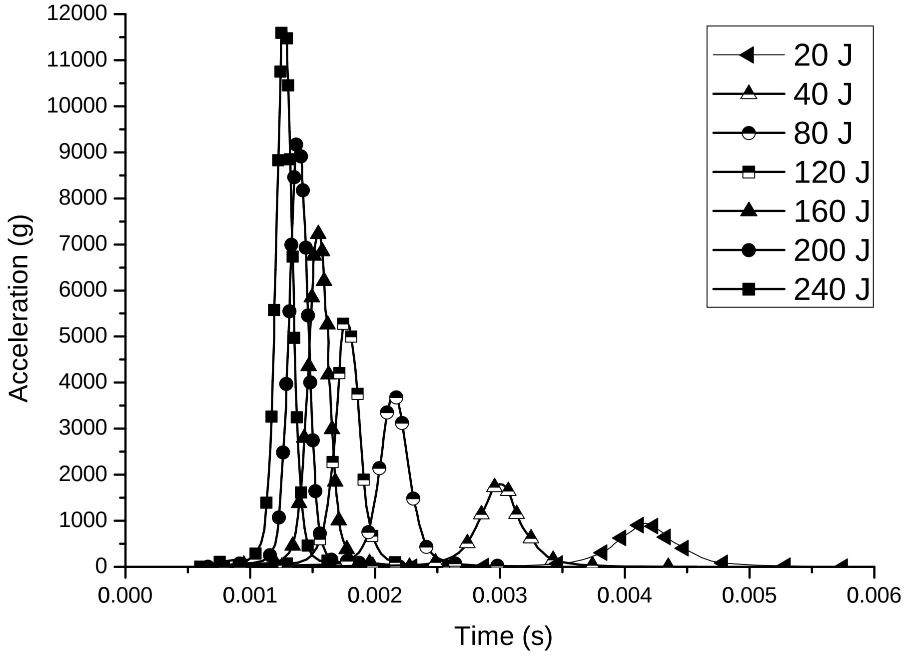


Figure 4.7: Headform impact test result for Full90

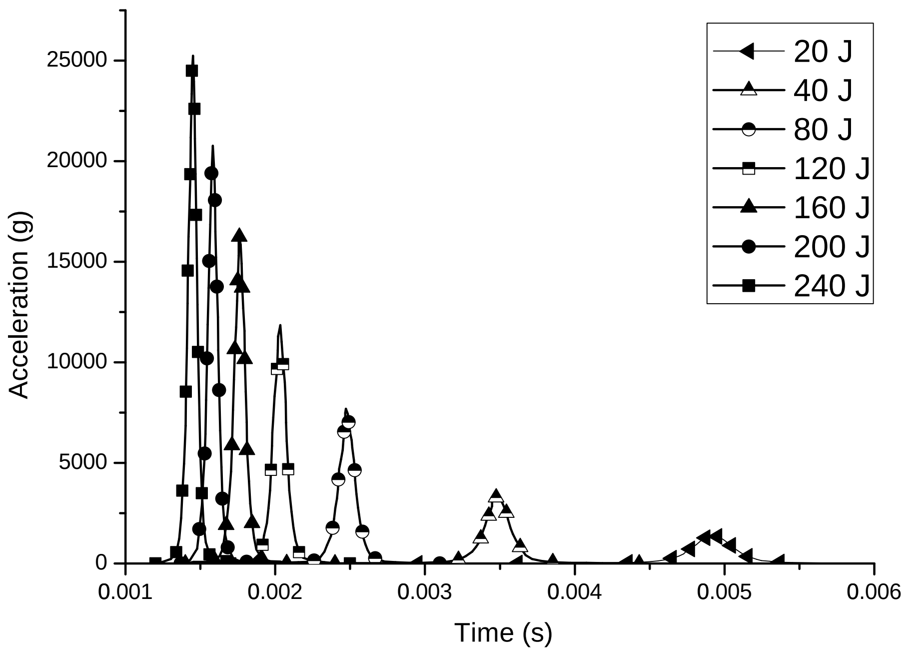


Figure 4.8: Headform impact test result for Force Field

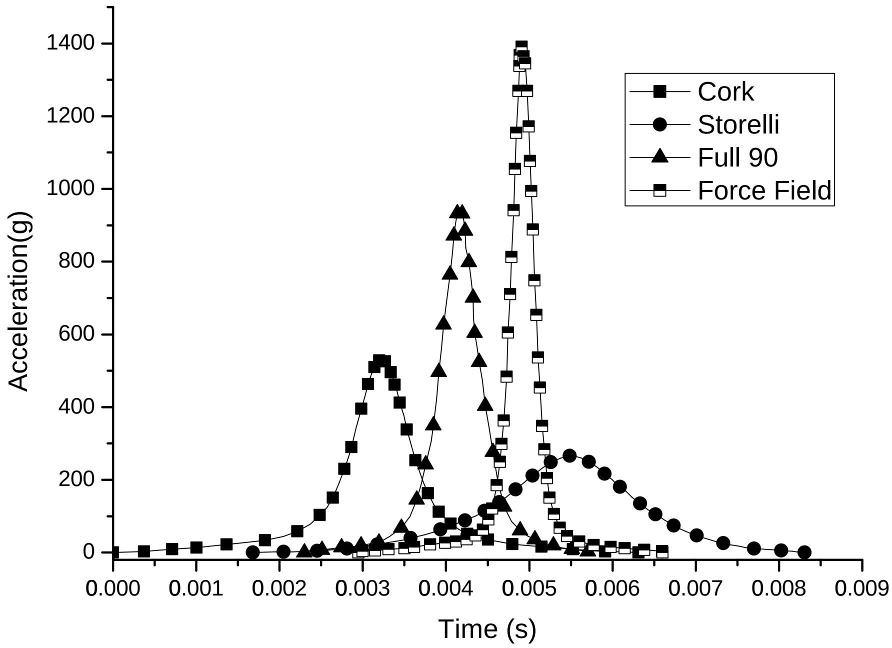


Figure 4.9: Headform impact test result for 20J

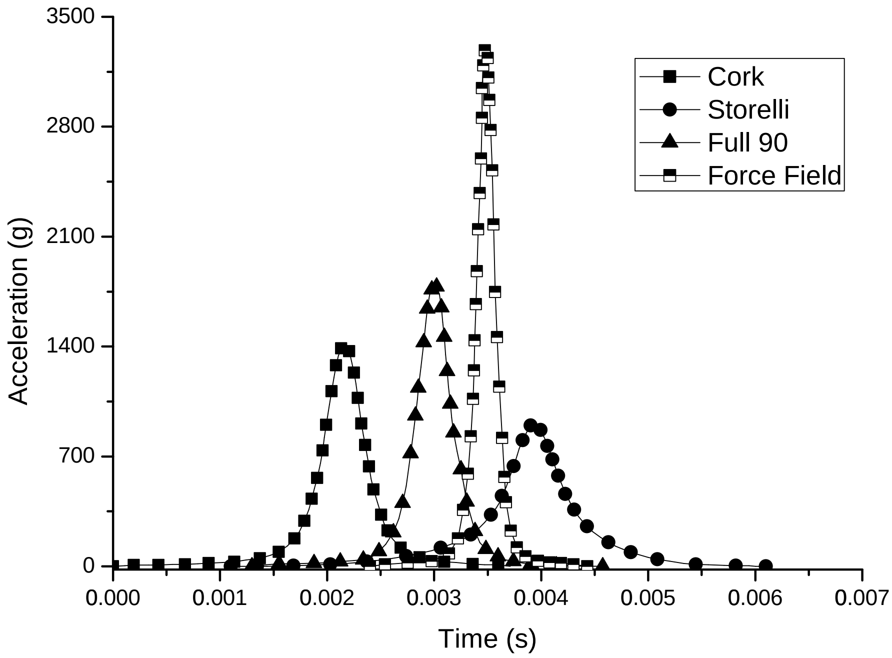


Figure 4.10: Headform impact test result for 40J

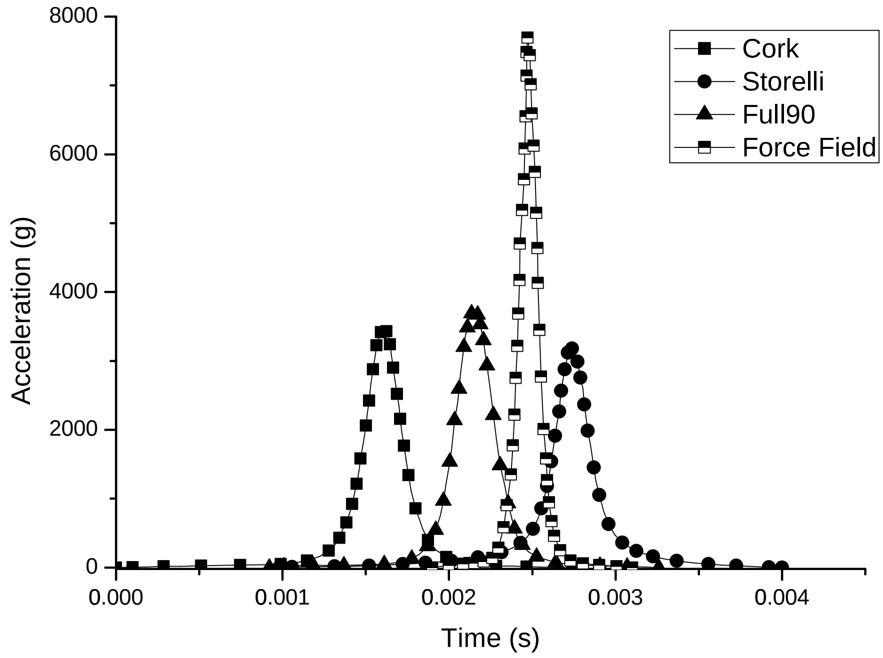


Figure 4.11: Headform impact test result for 80J

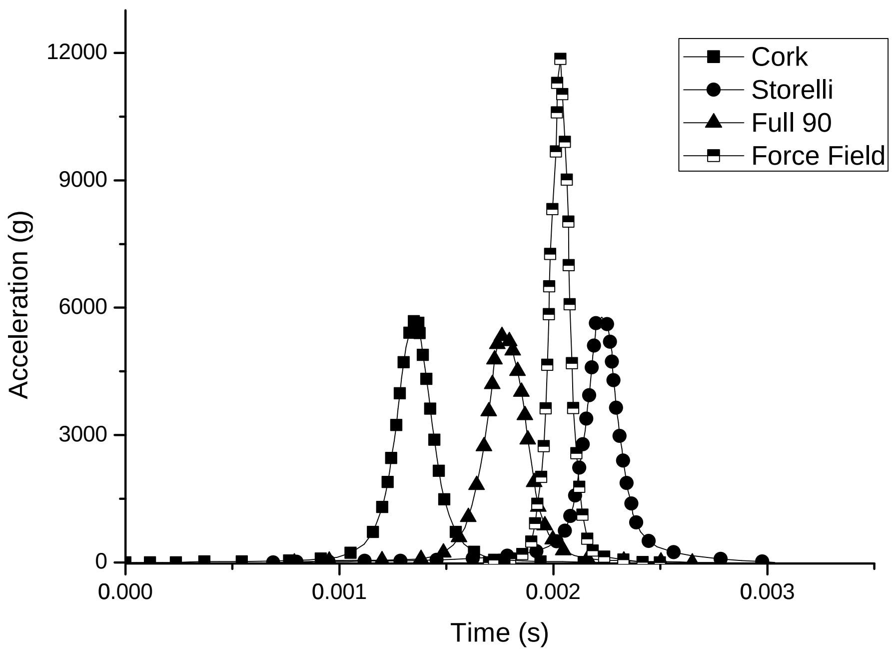


Figure 4.12: Headform impact test result for 120J

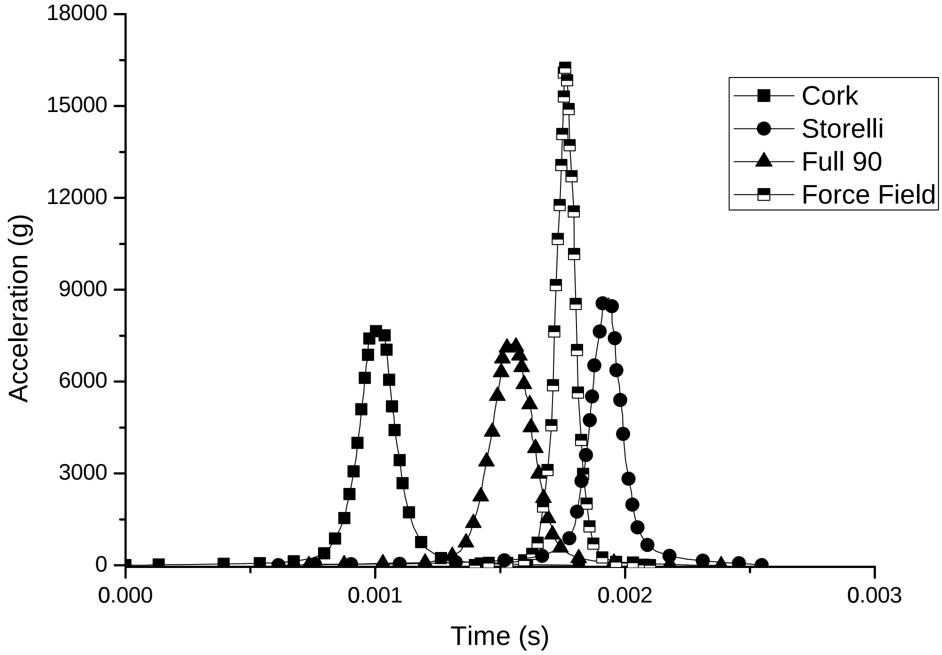


Figure 4.13: Headform impact test result for 160J

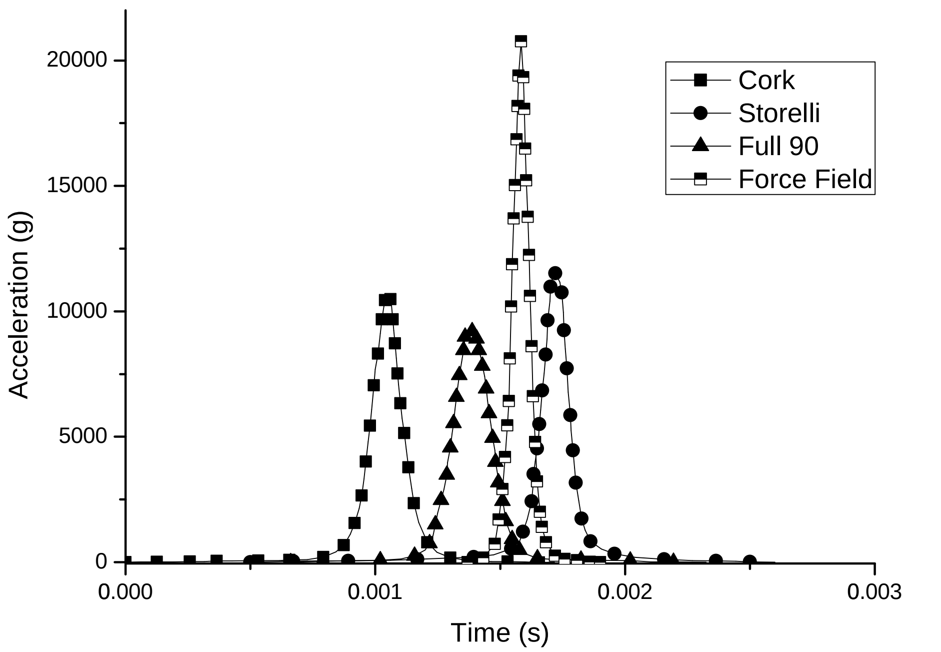


Figure 4.14: Headform impact test result for 200J

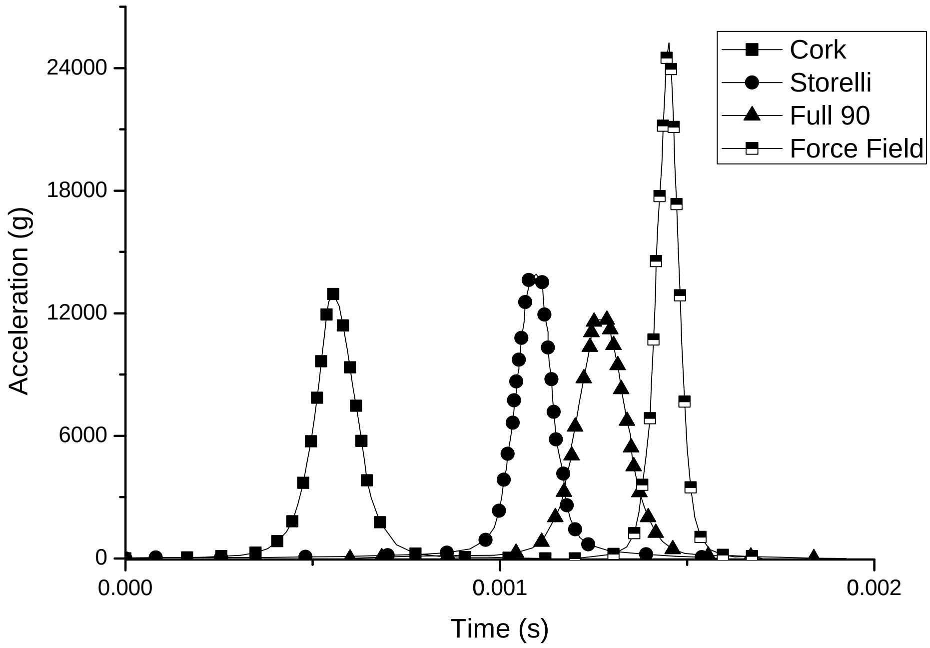


Figure 4.15: Headform impact test result for 240J

With these data it was possible to analysed the performance of all the materials in an impact situation. To do that it was used 3 head injury predictors: Peak of Linear Acceleration (PLA), Head Injury Criterion (HIC) and Wayne State Tolerance Curve (WSTC).

4.1.4 Peak of Linear Acceleration (PLA)

In the Figure 4.16 it is possible to see all the peaks of acceleration for each impact energy value and material.

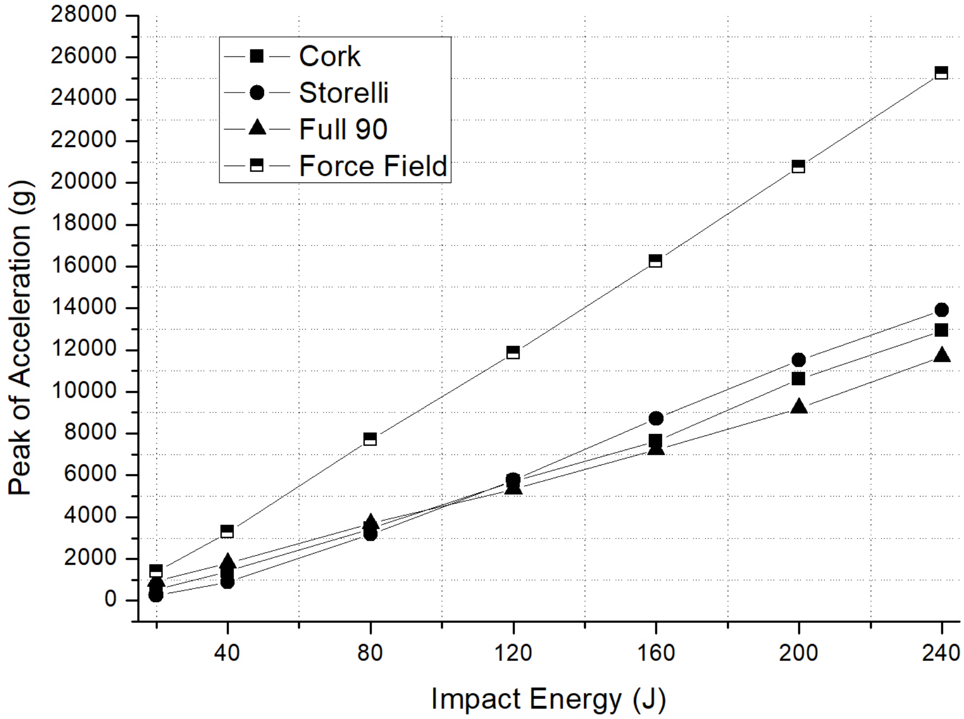


Figure 4.16: PLA results for each material in the Headform impact test

The first conclusion to make is about the Force Field foam because it had the worst results in terms of protection. The second is about the Storelli and the Full90 foams that changed its behaviour at 100 J, where the one with the best results changed to the worst between the last 3 materials.

The cork agglomerate like was observed maintains its results between the Storelli and the Full90 foam. Besides that, there is a energy level where it had the same level of protection of the synthetic foams.

Some authors use the PLA to stablish some thresholds creating a predictor related with head injuries. To identify some of them only the levels of 20 J and the 40 J were analysed because it was where there are values that do not exceed the thresholds. In the Table 4.4 it is presented all the results for the two energy values prior mentioned.

Table 4.4: PLA values for 20J and 40J

Material	20J [g]	40J [g]
Cork	530.46	1405.16
Storelli	266.33	900.00
Force Field	1390.55	3285.59
Full 90	941.96	1795.99

Comparing the prior results with the literature review it was concluded that all the values are higher than the thresholds except for the Storelli foam that only achieved to 50% of probability of a head critical injury (AIS 5). In terms of MTBI and Concussion all the materials achieved to its thresholds.

4.1.5 Head Injury Criterion (HIC)

In the Figure 4.18 it is possible to see all the HIC values and its behaviour for each material in all impact energy values.

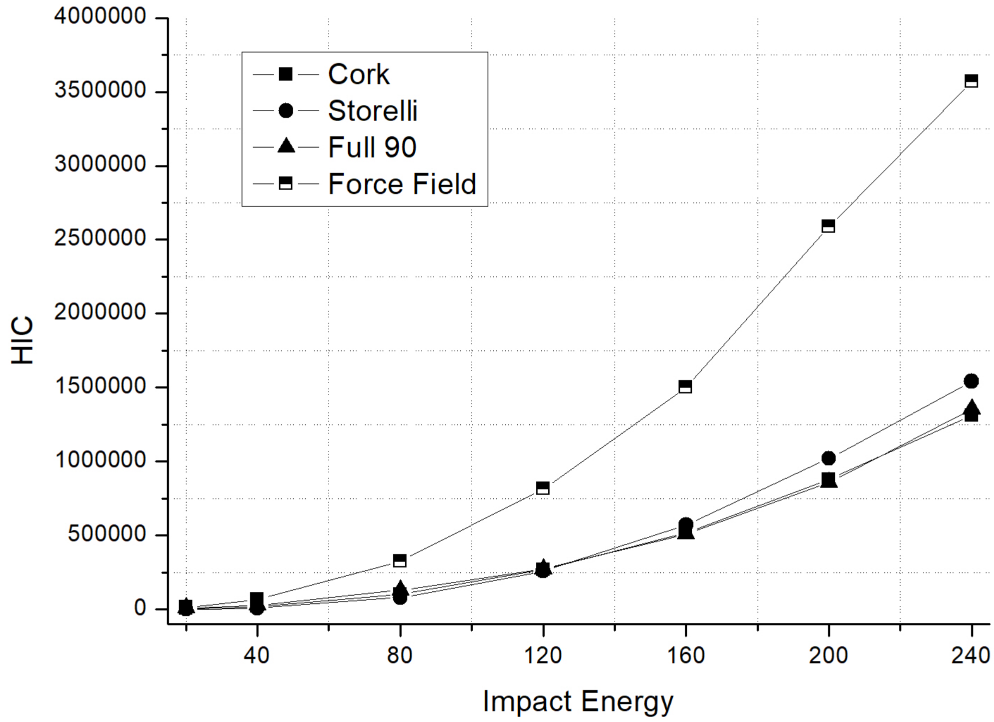


Figure 4.17: HIC results for each material in the headform impact test

For the Force Field foam, in this criterion, the conclusion is the same with the previous predictor, it has the worst behaviour. However, in this analysis until the 140J the other 3 materials have the same behaviour. After that, the Storelli results start to get worse but cork and Full 90 keep it almost the same.

In the literature, like with the previous predictor, has some thresholds that relate the values of HIC with the probability to occur some injuries. The energy levels analysed before were again 20J and 40J due to the magnitude of the results (Table 4.5).

Table 4.5: HIC values for 20J and 40J

Material	20J [g]	40J [g]
Cork	2893.10	18147.00
Storelli	1108.30	8053.70
Force Field	12666.00	66163.00
Full 90	7714.60	28135.00

Different from the first analysis this predictor has in consideration not just the peak of the linear acceleration but also its duration. This new variable change the results related to head injuries because now Force Field and Full 90 foams have bad results and surpass all the thresholds in the literature.

With better results, cork has values below to the threshold of HIC for 65% of probability of death and 99% of probability of life threatening injuries [138], [139].

As it was expected, the best results were from Storelli foam, that were below to the same thresholds of Cork but also to the 31% probability of death, to the 50% probability of a serious injury (AIS 3) and to the 50% probability of have a SDH.

However in terms of MTBI, severe TBI and Concussion not even cork or storelli foam were able to not exceed its threshold.

4.1.6 Wayne State Tolerance Curve (WSTC)

The final predictor was the WSTC that relate the peak of aceleration and the duration of it. These two variables were again taken from the 20J and 40J impact test (Table 4.6).

Table 4.6: Peak and pulse values for 20J and 40J

Material		20J	40J
Cork	Peak (g):	530.46	1405.16
	Pulse (ms):	6.31	3.71
Storelli	Peak (g):	266.33	900.00
	Pulse (ms):	6.63	5.00
Force Field	Peak (g):	1390.55	3285.59
	Pulse (ms):	3.65	2.03
Full 90	Peak (g):	941.96	1795.99
	Pulse (ms):	3.40	3.28

Once again just the cork and the Storelli result from the 20J were possible to analyse due to the peak not surpass the 600g, that is the limit acceleration of the WSTC.

The results show that these two material are within the area of injury (Figure 4.18), which means that exceed the human tolerance.

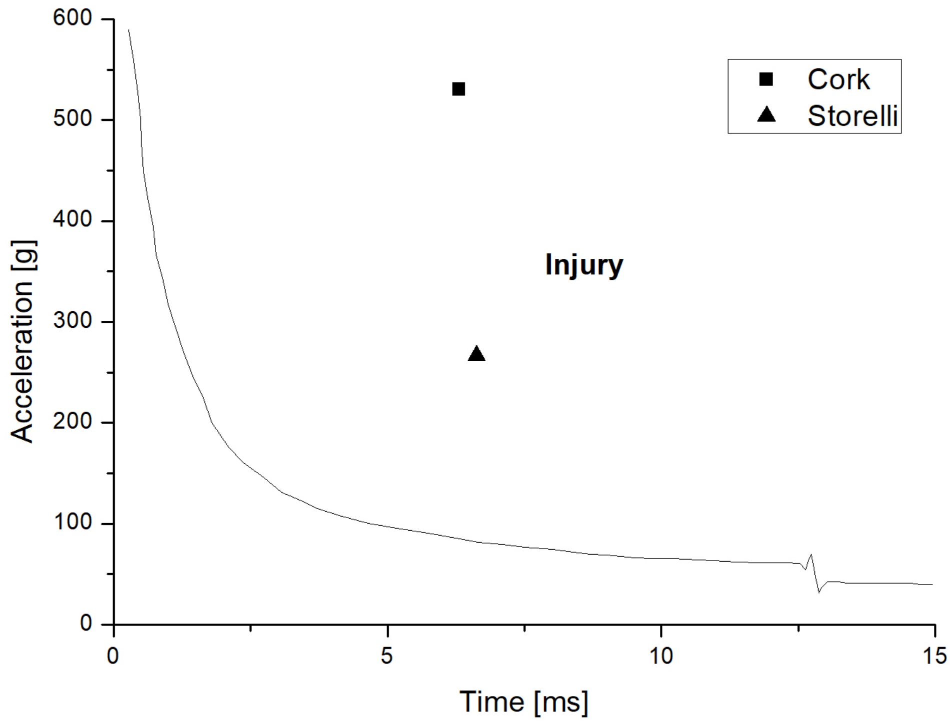


Figure 4.18: WSC results for Cork and Storelli in the Headform impact test

In the 3 analyses only some values could be related to head injuries because the headform test was over dimensioned. This happened due to the nature of the wall/ground (analytic rigid) that turn the impact more severe than in real life.

4.2 YEAHM tests

In this section it will be presented the brain behaviour to the impacts simulated in the previous test. To do that, it was used the YEAHM which motion and pressure response were based in the Nahum and Hardy experiments ([160], [181]) already used in some studies of the human brain [48]. In this case, the model had a new structure, the bringing veins, developed in a previous study [182].

4.2.1 Description of the YEAHM

The YEAHM (YEt Another Head Model) was based on medical images and is composed by skull, CSF and brain. In the Figure 4.19 is showed a cross section of the model and illustrates the anatomical features of the head. The brain model has all important sections: frontal, parietal, temporal, and occipital lobes, both hemispheres, cerebrum, cerebellum, corpus callosum, thalamus, midbrain, and brain stem. Membranes and bridging veins are included in CSF part.

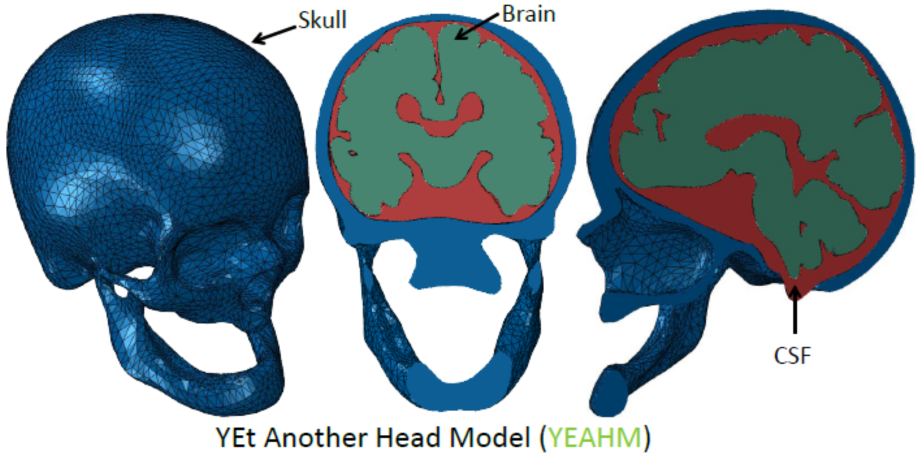


Figure 4.19: Yeahm model [48].

The viability of the model is most due to the material formulation. In the brain it was used hyperelastic and viscoelastic material, in the CSF it was hyperelastic and in the skull elastic. For all these parts was defined a density as well. All these parameters are presented in the Table 4.7.

Table 4.7: Material properties of YEAHM model

Brain							
$\rho[kg/m^3]$	$\mu[MPa]$	α_1	$D_1[MPa^{-1}]$	g_1	g_2	$\tau_1[s]$	$\tau_2[s]$
1040	0.012	5.0507	0.04	0.5837	0.2387	0.02571	0.02570

CSF			
$\rho[kg/m^3]$	C_{10}	C_{01}	$D_1[MPa^{-1}]$
1000	0.9	1.0	0.9

Skull		
$\rho[kg/m^3]$	$E[MPa]$	ν
1800	6000	0.21

The material properties presented in the previous table are: ρ - density; E - Young's modulus; ν - Poisson's ratio; μ - shear modulus; g - relaxation coefficients; τ - relaxation time and C_{10}, C_{01}, α_1, D_1 represent some material parameters.

In terms of boundary conditions and contact algorithms, this model use finite-sliding formulation and kinematic contact between the CSF and the brain and between the CSF and the skull with a friction coefficient of 0.2 for tangential behaviour [48].

In the mesh definition of this model was used 836328 elements for the brain, 57257 elements for the skull and 98032 elements for the CSF. The type of elements used is the second order C3D10M, that works well in contact situations and is robust in volumetric locking or transverse shear one. When the hourglass control (M) is activated, normally this problem does not propagate [182].

However, like was mentioned before, in this study the YEAHM has an new structure, the bridging veins.

4.2.2 Bridging veins (BV)

The bridging veins have the function of draining the venous blood from the cerebral cortex to the superior sagittal sinus (SSS) [183] crossing the subdural space. The distributions of all the BV connections through the SSS are not uniform. Thus, the flow has a range of directions and because of that any brain movement in relation to the skull will result in normal and shear loads. That event could result in the rupture of the BV and consequently the possibility of injury [184], like SDH.

4.2.3 YEAHM and BV model

If it was used just the bridging veins in the model would be a limit method because the connection between the brain and the skull would be the same of the YEAHM. Thus, this new model includes the SSS, (Figure 4.20), to give a more realistic behaviour of the bridging veins in all the directions but also to be more precise in the SDH detection.

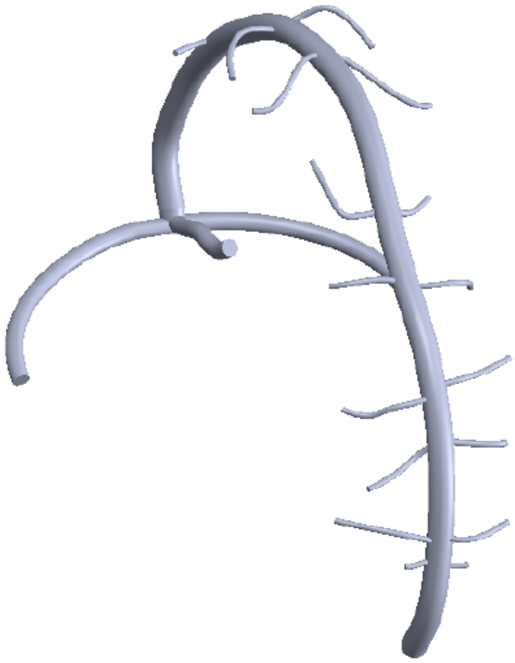


Figure 4.20: Bridging veins and superior sagital sinus model [182].

Figure 4.21 presents the assembly used for the impact tests.

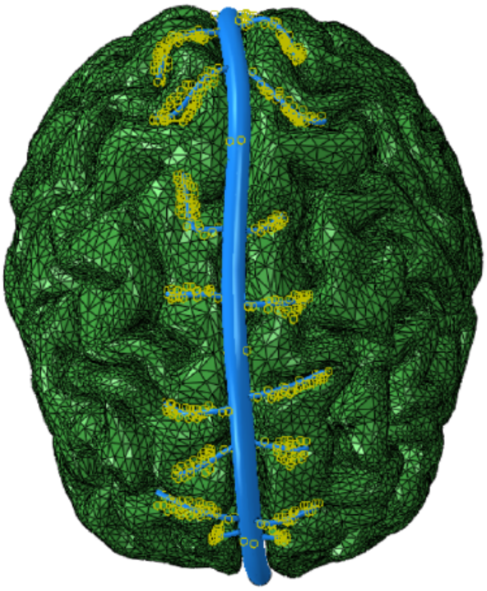


Figure 4.21: YEAHM with BV [182].

4.2.4 Test and results

The next step was to insert the 3 linear acceleration components from the headform test of 20J and 40J (Tables 4.5 - 4.8) in the new model center of mass to see the brain response in that impacts events.

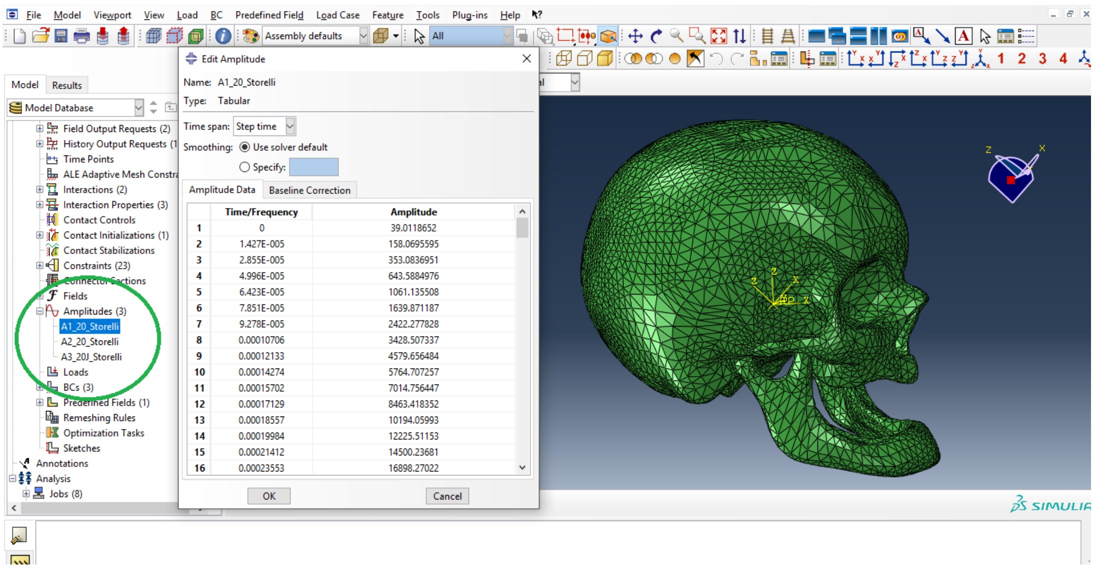


Figure 4.22: Acceleration curve inserted in Abaqus.

The data from the simulation was analysed following 4 parameters: strain, pressure, von Mises stress and the rupture of the bridging veins. With the thresholds already presented in the literature review, was possible to relate the results with the probability to occur MTBI, DAI and Concussion. Some of these reference values refer to a single area of the brain like the corpus callosum (Figure 4.23), where is possible to analyse the 50% probability to occur Concussion and DAI.

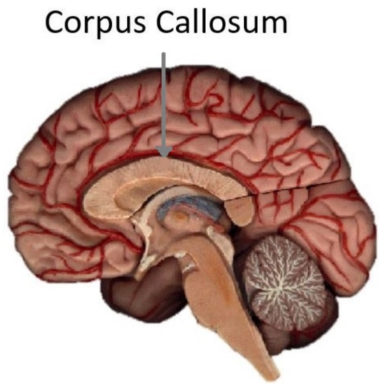


Figure 4.23: Corpus Callosum [185].

To relate and compare the results of 2 moments of the impact test were considered, the coup, the site of impact and the contrecoup, the opposite site of the impact [48], (Figure 4.24).

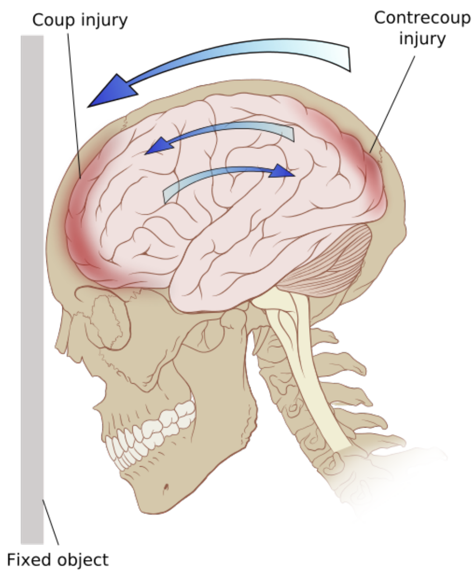


Figure 4.24: Coup and Contrecoup [186].

The final results of these brain impact tests are presented in the following tables. The representation of it consists in a rainbow when the values are below the thresholds and white when are above. Only in the brain pressure analysis it was stablish a minimum and in that case the values below are represented in dark blue. The first ones are the von Mises stress results where was consider 50% probability to occur MTBI.

The second set of results are related to the brain pressure where was used the thresholds from moderate and sever TBI and established as minimum and maximum.

The third set of results are related to the strain considered DAI, 50% probability to occur DAI and Concussion in the corpus callosum.

The last ones refer to the BV rupture that are consequently related with SDH. In the Figure 4.25, there is a representation of a BV rupture in the simulation that helped to build the Table 4.18 with the final results in this issue.

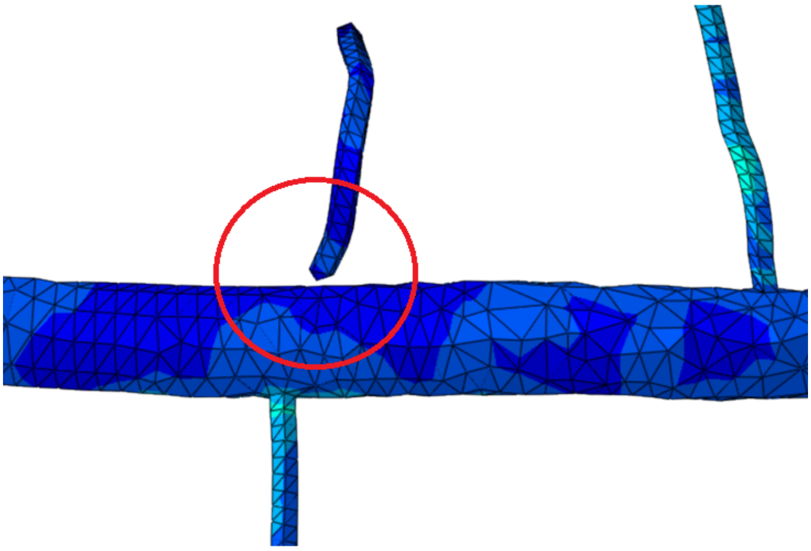


Figure 4.25: BV rupture in FEHM

Every result are presented in the coup situation but in some cases the contrecoup was Not Available (NA), which means that this moment wasn't observed in the simulation.

Table 4.8: von Mises stresses related to the probability of 50% to have MTBI (18 kPa) in the brain test with 20J.

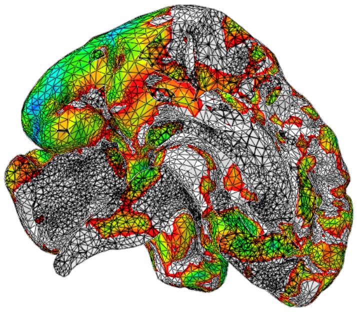
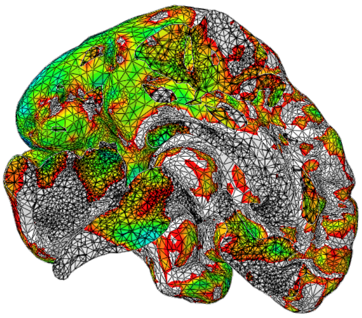
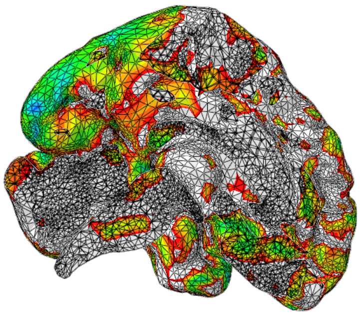
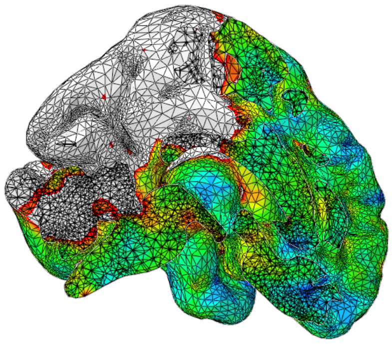
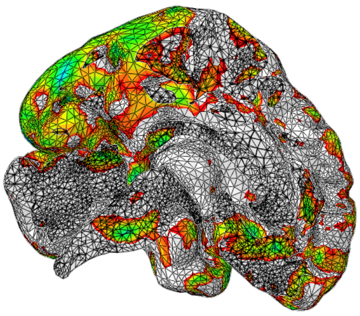
Material	**Coup**	**Contrecoup**
Cork		NA
Storelli		NA
Full 90		
Force Field		NA

Table 4.9: von Mises stresses related to the probability of 50% to have MTBI (18 kPa) in the brain test with 40J.

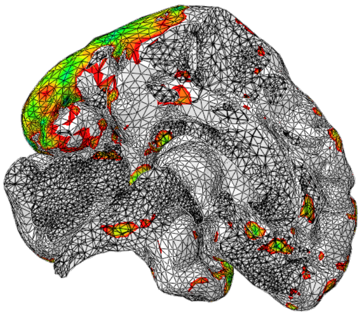
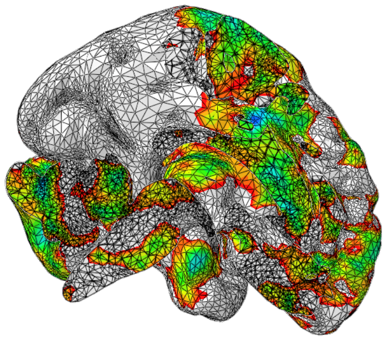
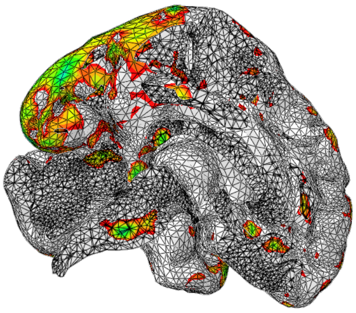
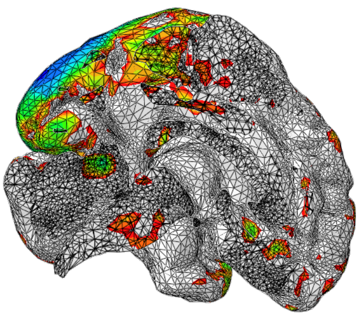
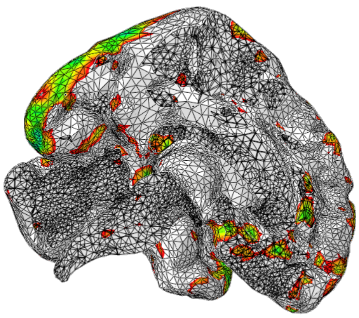
Material	**Coup**	**Contrecoup**
Cork		
Storelli		NA
Full 90		NA
Force Field		NA

Table 4.10: Pressure related to Moderate and Severe TBI (minimum 173kPa and maximum 235kPa) in the brain test with 20J.

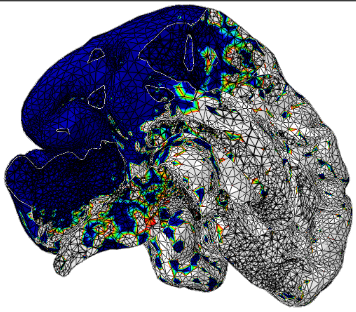
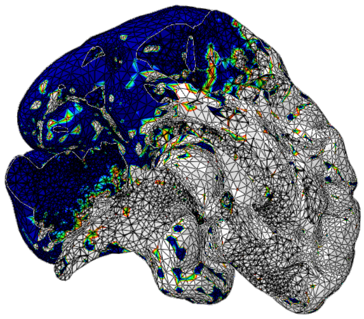
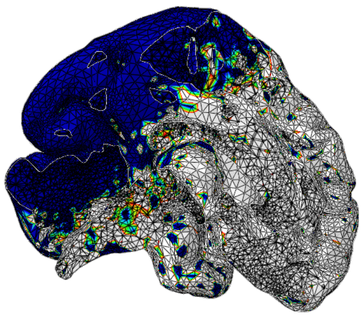
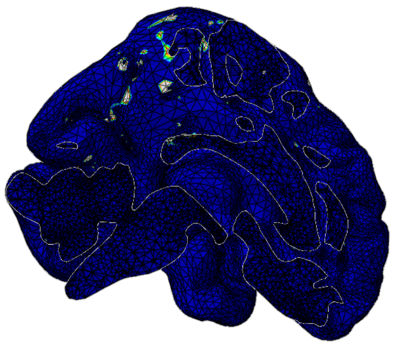
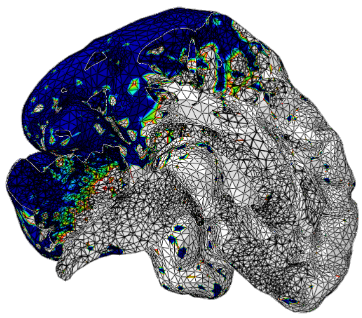
Material	**Coup**	**Contrecoup**
Cork		NA
Storelli		NA
Full 90		
Force Field		NA

Table 4.11: Pressure related to Moderate and Severe TBI (minimum 173kPa and maximum 235kPa) in the brain test with 40J.

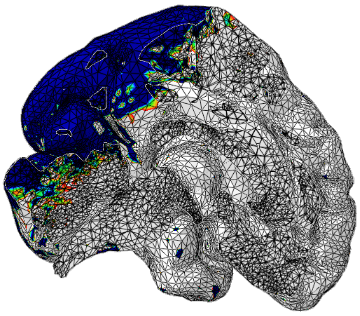
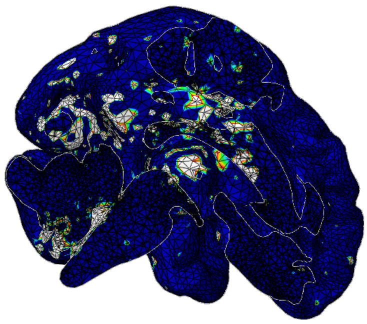
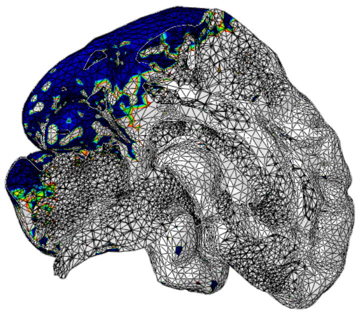
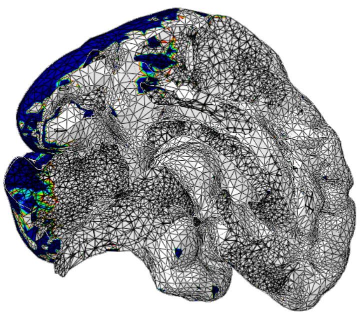
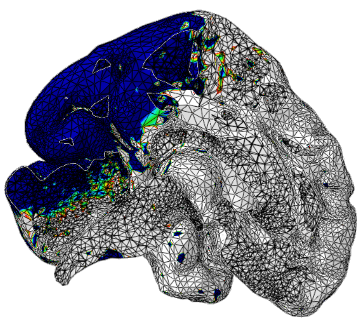
Material	**Coup**	**Contrecoup**
Cork		
Storelli		NA
Full 90		NA
Force Field		NA

Table 4.12: Maximum principal strain related to DAI (0.18) in the brain test with 20J.

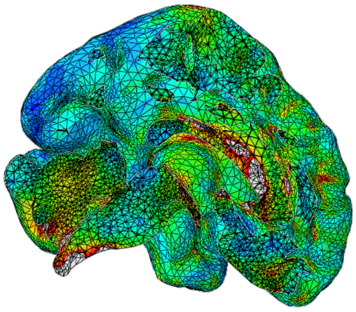
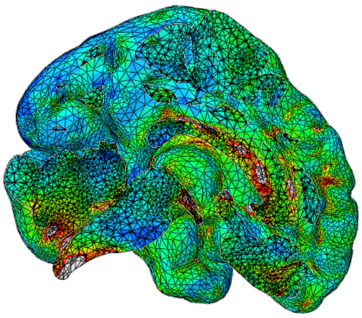
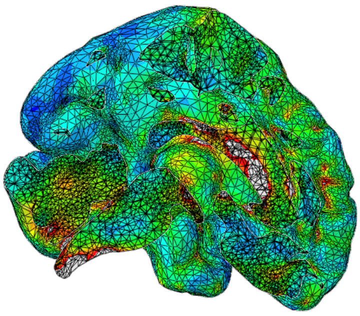
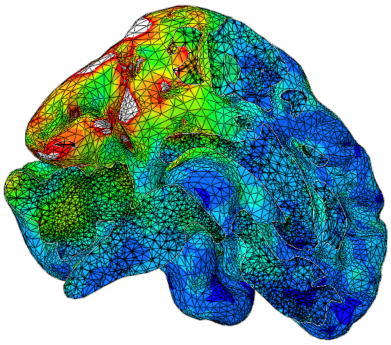
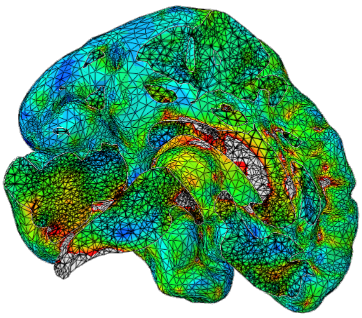
Material	Coup	Contrecoup
Cork		NA
Storelli		NA
Full 90		
Force Field		NA

Table 4.13: Maximum principal strain related to DAI (0.18) in the brain test with 40J.

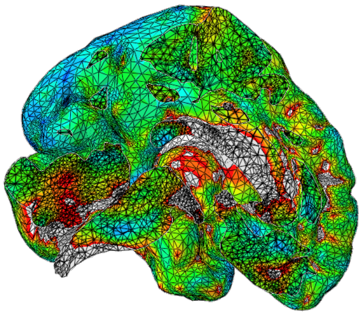
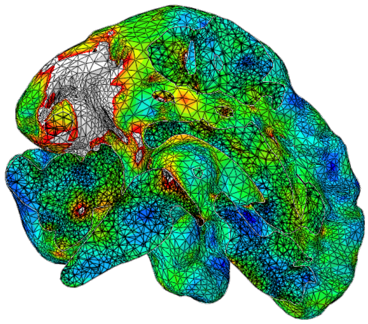
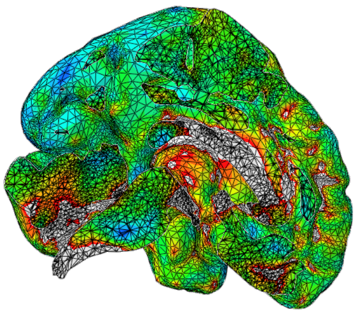
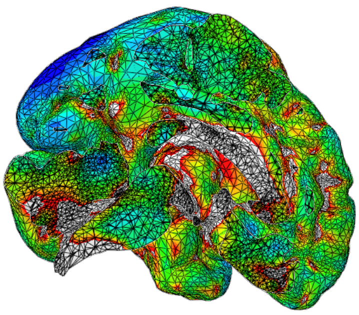
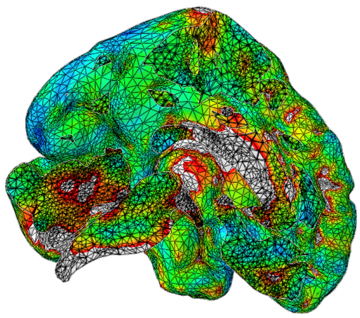
Material	**Coup**	**Contrecoup**
Cork		
Storelli		NA
Full 90		NA
Force Field		NA

Table 4.14: Maximum principal strain related to the probability of 50% to have DAI in the corpus callosum (0.21), in the brain test with 20J.

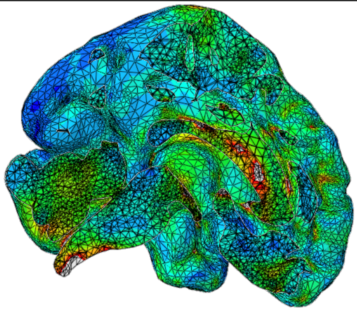
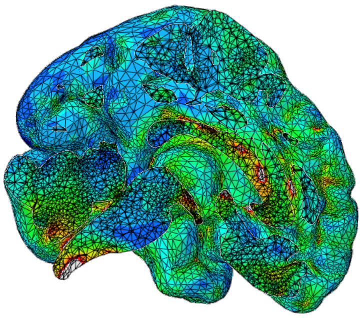
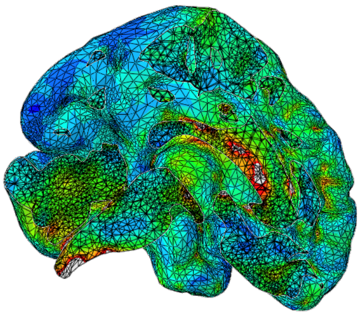
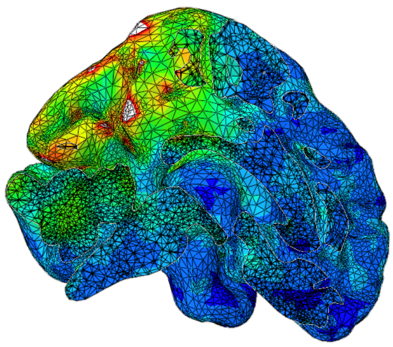
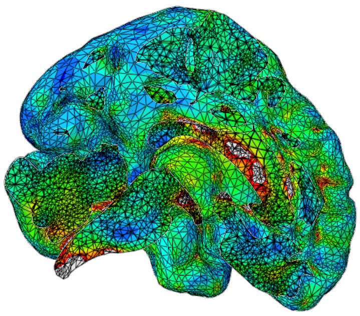
Material	**Coup**	**Contrecoup**
Cork		NA
Storelli		NA
Full 90		
Force Field		NA

Table 4.15: Maximum principal strain related to the probability of 50% to have DAI in the corpus callosum (0.21), in the brain test with 40J.

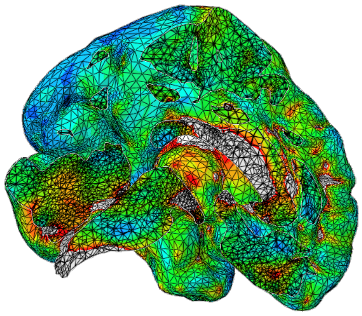
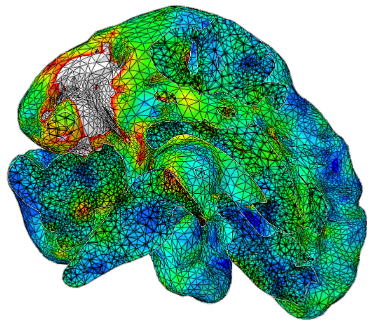
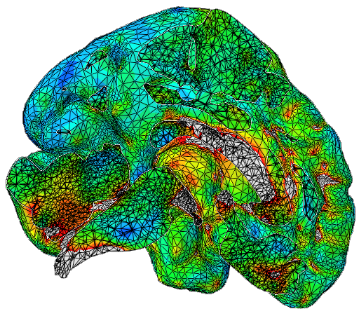
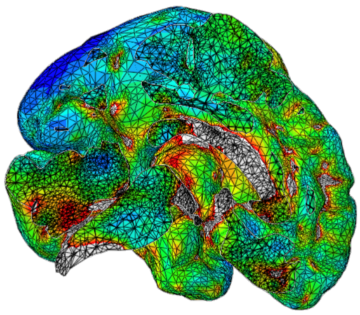
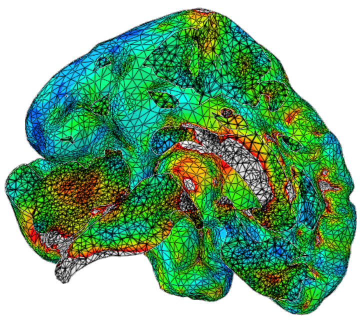
Material	**Coup**	**Contrecoup**
Cork		
Storelli		NA
Full 90		NA
Force Field		NA

Table 4.16: Maximum principal strain related to the probability of 50% to have Concussion in the corpus callosum (0.15), in the brain test with 20J.

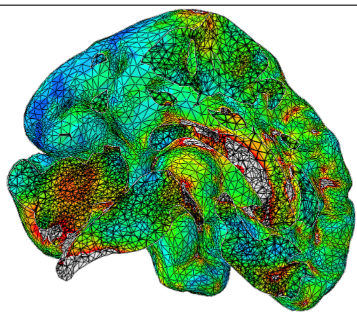
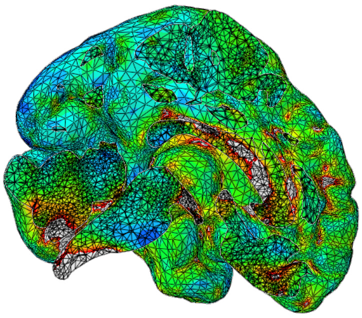
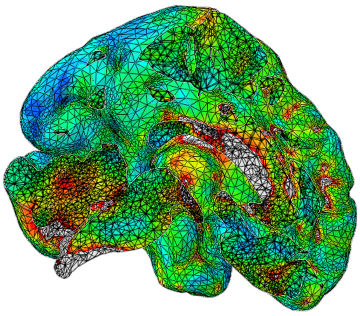
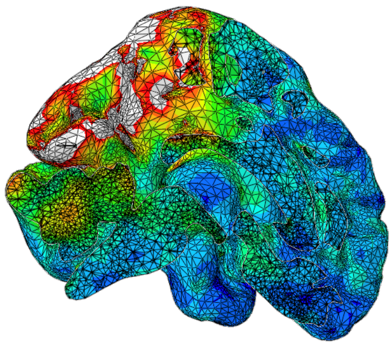
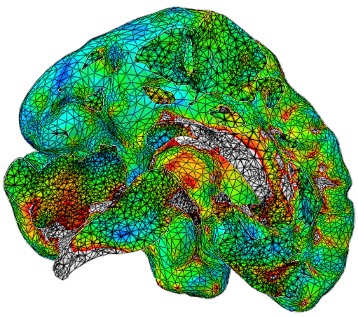
Material	**Coup**	**Contrecoup**
Cork		NA
Storelli		NA
Full 90		
Force Field		NA

Table 4.17: Maximum principal strain related to the probability of 50% to have Concussion in the corpus callosum (0.15), in the brain test with 40J.

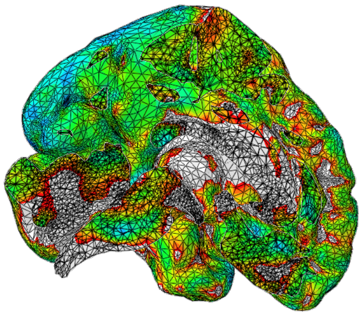
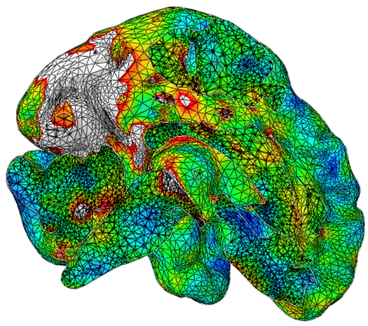
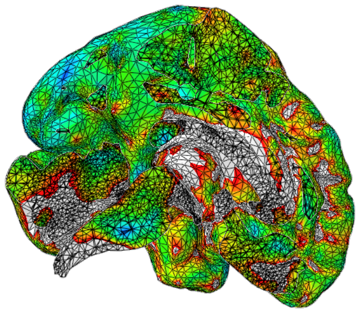
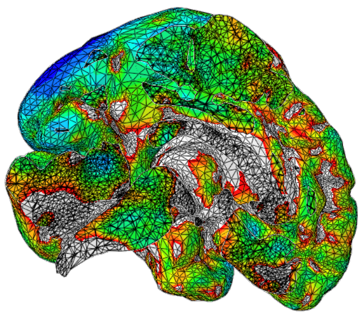
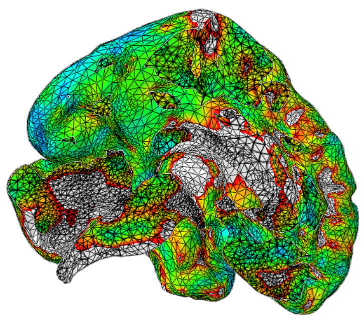
Material	**Coup**	**Contrecoup**
Cork		
Storelli		NA
Full 90		NA
Force Field		NA

Table 4.18: Verification of bridging veins rupture and consequently SDH.

Material	Bridging Veins Rupture
Cork:	
20J	Yes
40J	Yes
Storelli:	
20J	No
40J	Yes
Full 90:	
20J	Yes
40J	Yes
Force Field:	
20J	Yes
40J	Yes

As it was expected, these results validated the conclusions from the previous section where all the values are above the thresholds from the literature. The section view of the model show that behaviour too. The white zone that appear in all the results but in terms of the maximum principal strain the area was much small. In terms of BV rupture only the Storelli test with 20J was capable of maintain the veins safe.

In the material evaluation the results were the same too because agglomerate cork and Storelli headband had the best response to the impact.

In general the results were not so good due to the fact already mentioned about the oversized impact results.

Chapter 5

Conclusions and future works

This chapter presents the general and main conclusions, and discuss the results obtained in this work. In addition, some research ideas that may be implemented in related future works are suggested as well as some acquired competences.

5.1 Conclusions

Nowadays, the number of players in contact-sports have been growing as well as the young players that start playing sooner. This fact brought a new worry into the sport world in terms of head injuries like Concussion, CTE, DAI and SDH, and consequently the protective equipment starts to be improved in order to solve this problem.

The majority of these devices is made of synthetic foams and able to absorb reasonable amounts of energy although, some of them do it by deforming permanently.

In a society continuously searching for new environmentally friendly and sustainable resources, a material such as cork can be a natural alternative to synthetic materials. Cork is a natural cellular material capable of absorbing great amounts of energy. In addition, cork recovers almost entirely after deformation, which is a desirable characteristic in multi-impact applications. Besides that, agglomerated cork can be produced with the density that the manufacturer want which increase the range of applications to this material.

This is a good alternative for devices like headbands. It is used in sports where the use of head protection it is not required but the players want to feel safer without decrease their performance.

In order to assess the applicability of agglomerated cork as energy absorber in a headband and see its potential against the ones in the market, several steps were performed. First, experimental tests were performed on different types of agglomerated cork and 3 headbands from the market in order to characterise these materials. Quasi-static and dynamic tests were performed to select the most promising cork agglomerates. This step ended with the numerical validation of that tests in order to validate the constitutive laws of each material.

The final step consisted in the creation of a FE headband model with the help of a ECE 22.05 headform model. This system was used in an impact test with different energy impact values to see the potential of all the materials in such situations. As the wall/ground where the headform hit is analytic rigid the impact became more severe than in a real situation, which resulted in overestimated values. However, the comparison between materials could still be possible.

Finally, the curves from the headform center of mass were used to drive YEAHM, with the BV model, in order to analyse the helmet from a biomechanical point of view. Concussion, DAI and MTBI thresholds were used in order to access to these injuries in the impacts. As in the previous one it was possible to compare the material, although the results in terms of head injuries were overestimated.

Thus, it was concluded that agglomerated cork NL10, can compete with the material from the headbands at the market and in some cases it has better response to impacts. However, there is a lot to do in this area because with a parametric study and more accurate simulation of the real situations in sports, the best cork-headband with the best protection and thinner layers could be created.

Cork application is not limited to headbands and has the potential to be applied in other types of personal safety gear or even in other applications where its characteristics are desirable. In a society constantly looking for natural and sustainable resources, change synthetic material for a natural material like cork is a good solution.

5.2 Competences acquired

With the development of this thesis there were some new competences acquired in different areas:

- Simulation;
- Experimental material test analysis;
- Entrepreneurship.

Work in explicit simulation with the material formulations from Abaqus software were two new challenges.

The same happen in the experimental test because the material behaviour was different from the ones I was used to work. Cellular materials have a different response to compression forces comparing to metals.

The last area mentioned is related to a course in entrepreneurship from the Erasmus+ organized within the framework of the project ABC-MELES 2.0. It helped to know if the cork-headband was good for the market and its public target.

5.3 Future works

Considering the conclusions previously presented, the following future work is suggested:

- Parametric study of a headband for soccer.
- Development of a cork-headgear for in-game recovery.
- Cork application in skateboard protection equipment.

The first topic it's the second step of the work developed in this thesis where the simulation will recreate real situation in a specific sport.

The second suggestion is about an application already made with headgear, in-game recovery. The incorporation of cork in this protective device could turn it thinner and also more comfortable for the user.

The third is about the cork application in impact situations more severe that as was concluded in this thesis has a good response do that.

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